Plain Film Tomography

1) Problem of Superposition

- A planar X-ray image is a 2D representation of a 3D object.

- Consequently structures are positioned above each other and can be hard to distinguish.

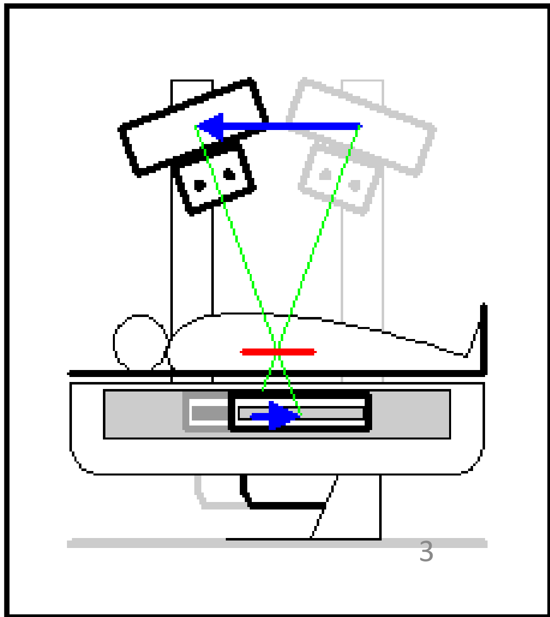
Linear Tomography

- **2) Goal**
  - keep plane of interest in focus
  - blur all other plans
    - enhances contrast

- **3) Idea:**

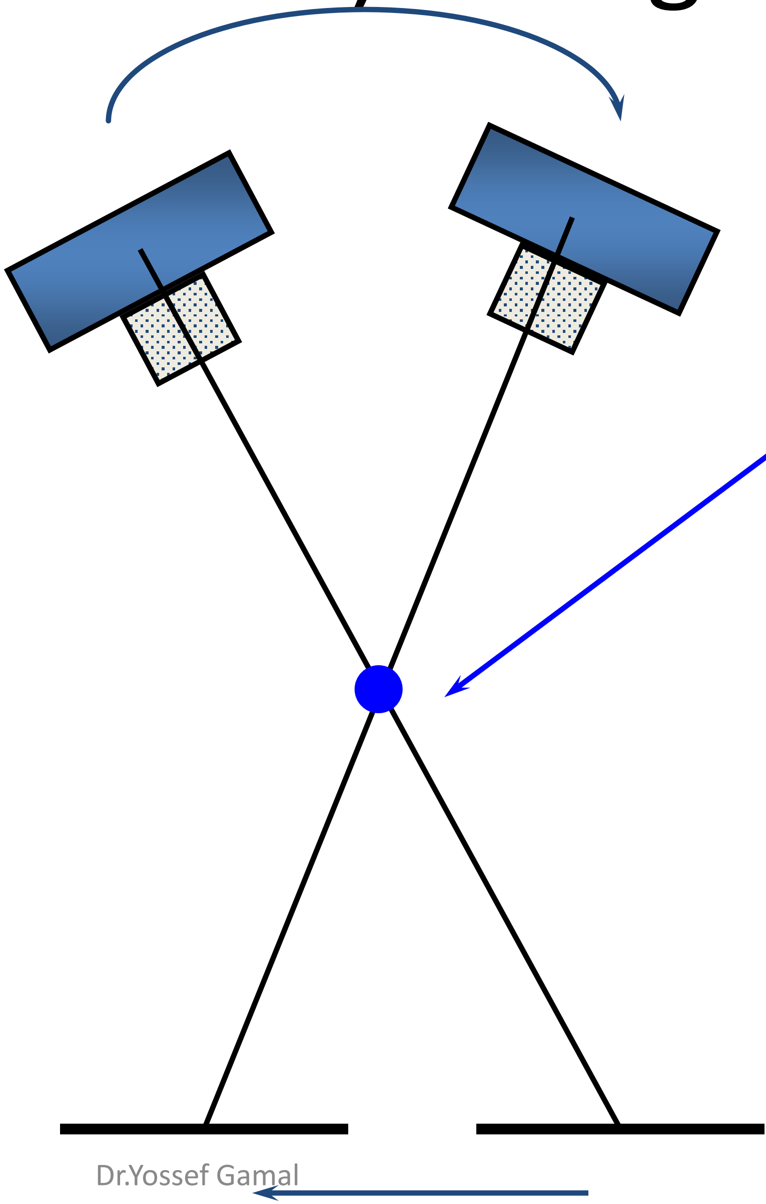
-System composed of tube , film and patient

-Blurring accomplished by synchronous movement of two of the components (usually tube & film)



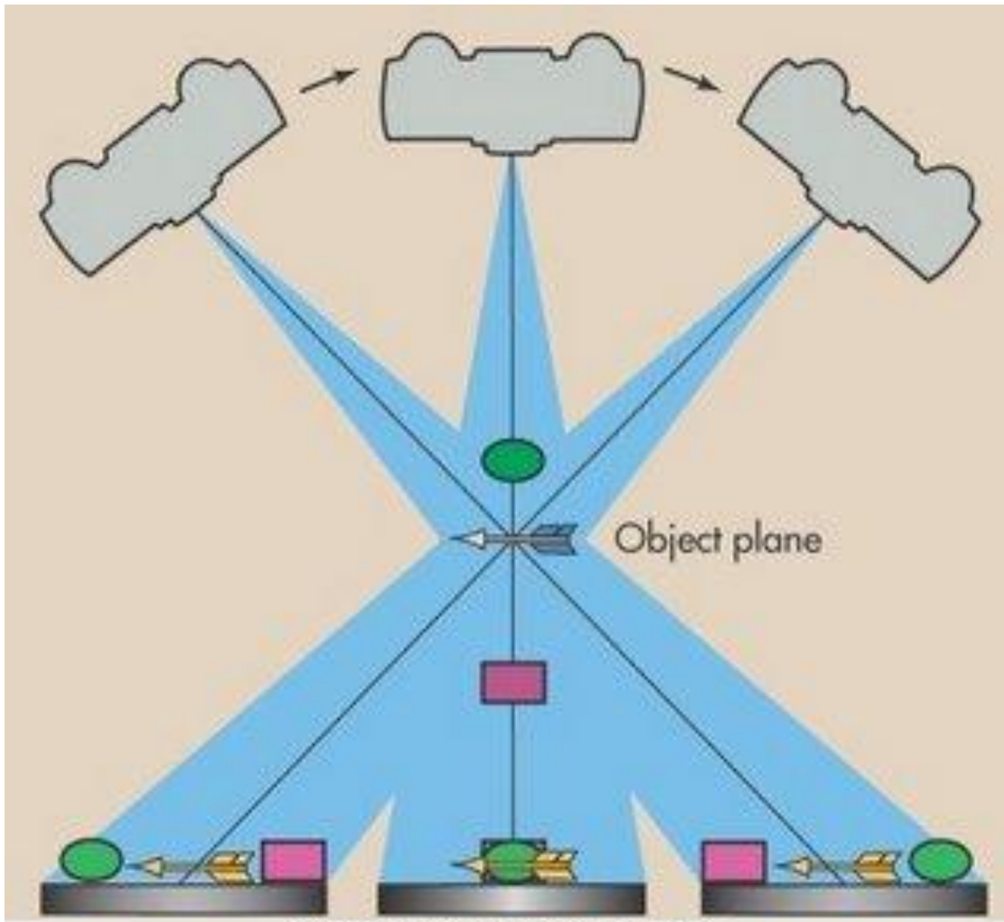
- Popularity decreasing because of
  - CT
  - MRI

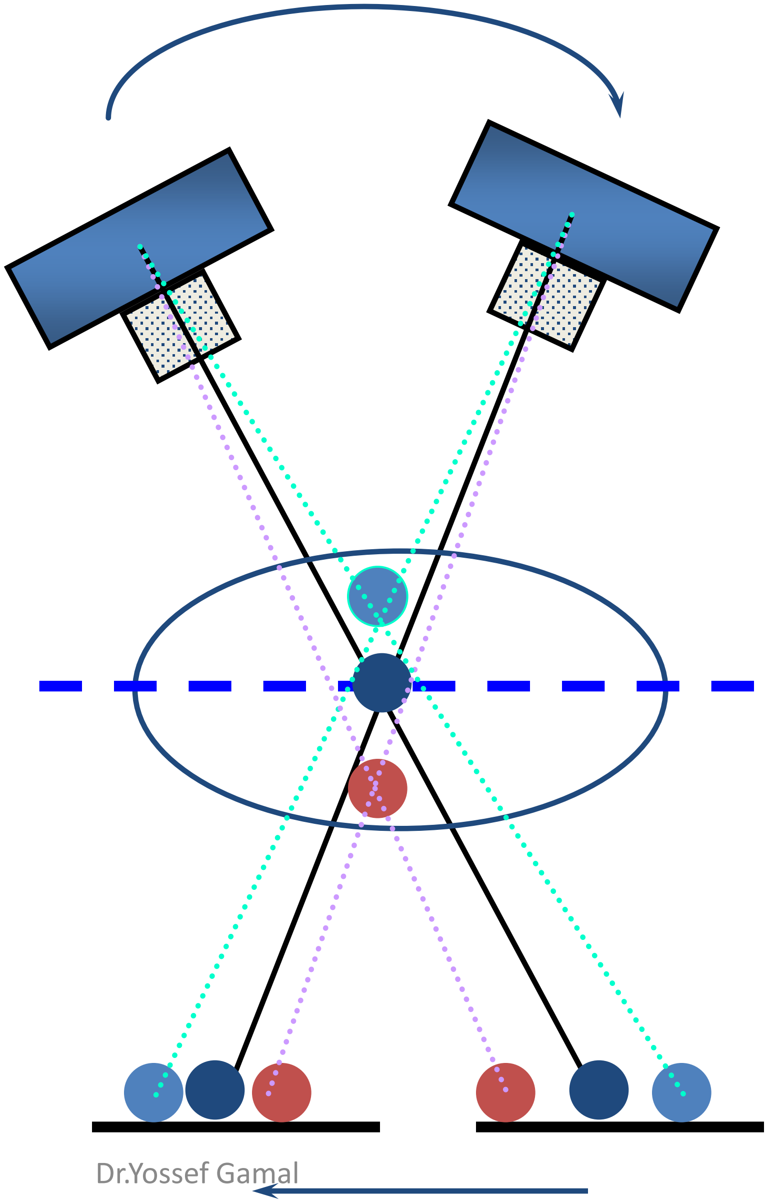
4)Tomography components



- tube & bucky connected by rod mechanically (or electronically)
- rod hinged around fulcrum
- tube moves horizontally in one direction
- film moves in other direction
- Tube head is rotating to make central ray always directed to the pivot point

5) Blurring mechanism



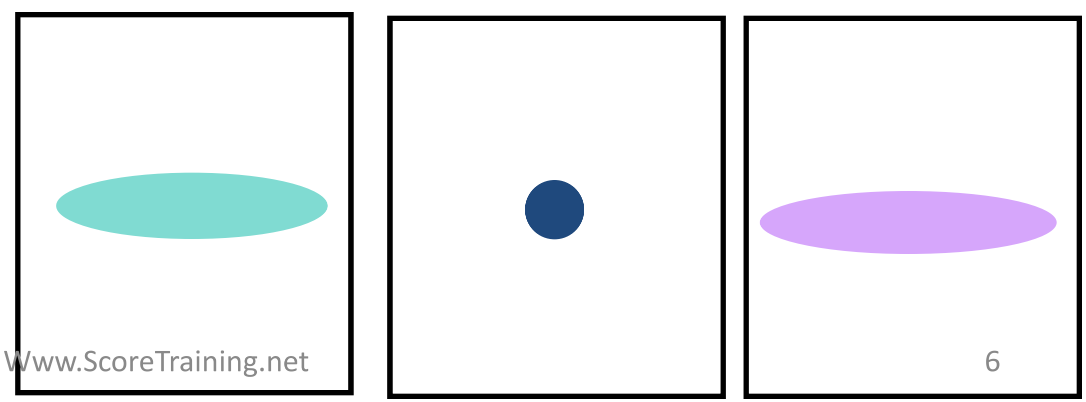


- **Objects above or below fulcrum plane:**

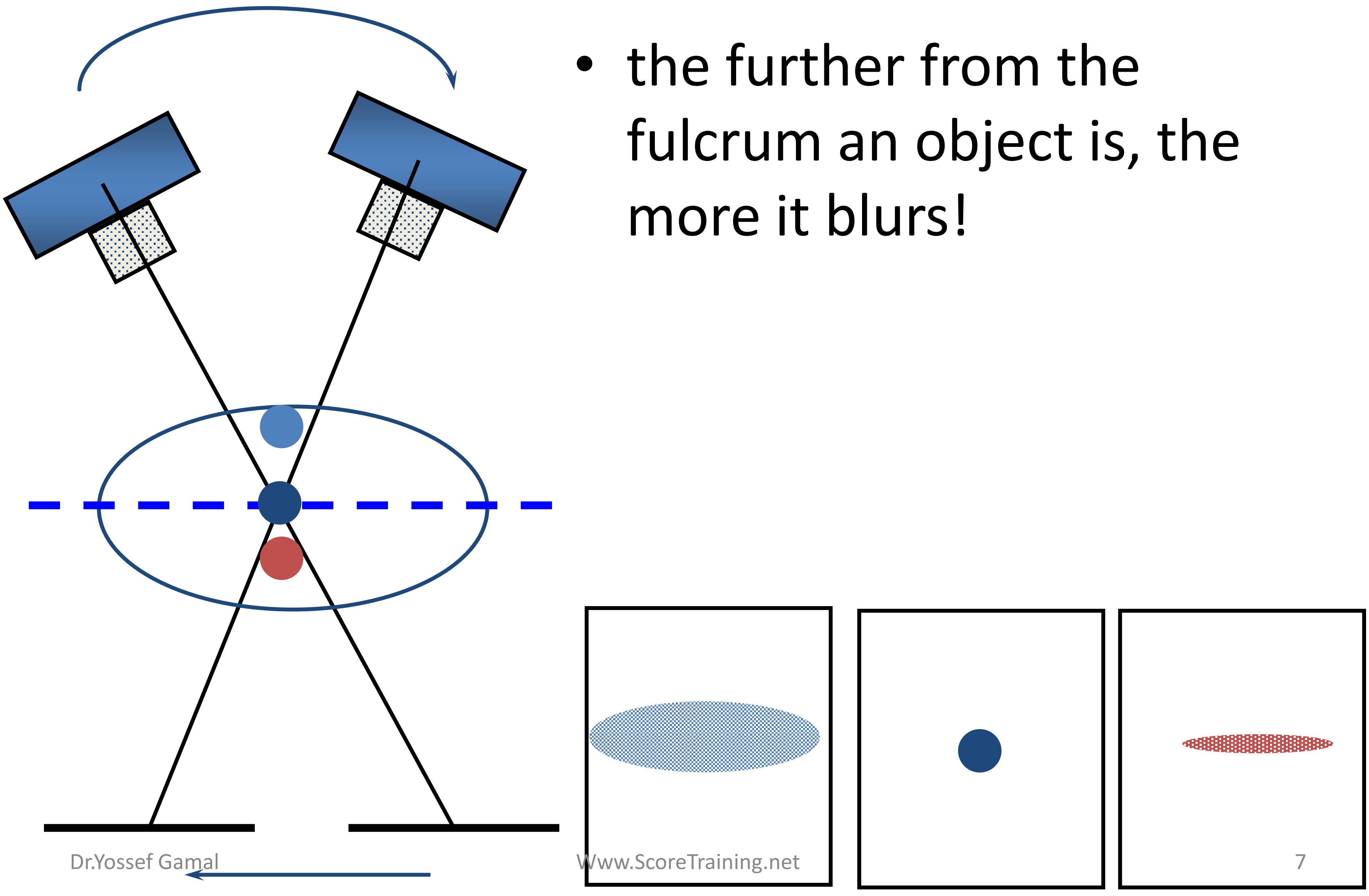
These objects change position on film

i.e. shadows on the film move different distance than moved by the film

Thus blur

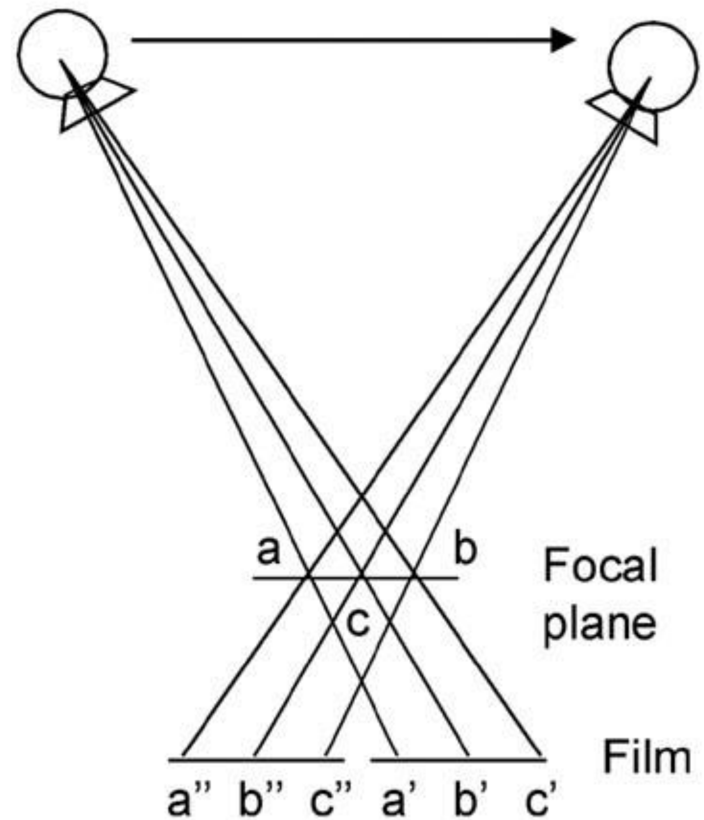


- the further from the fulcrum an object is, the more it blurs!



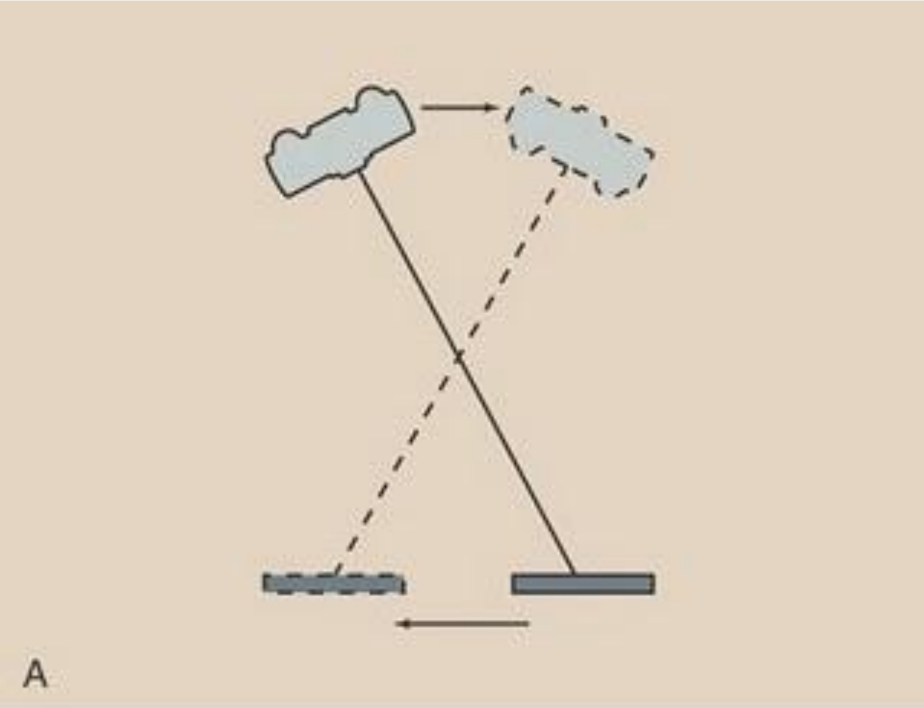
- **Objects at the level of the fulcrum:**

Their shadows on the film move at the same speed as the cassette → sharp image

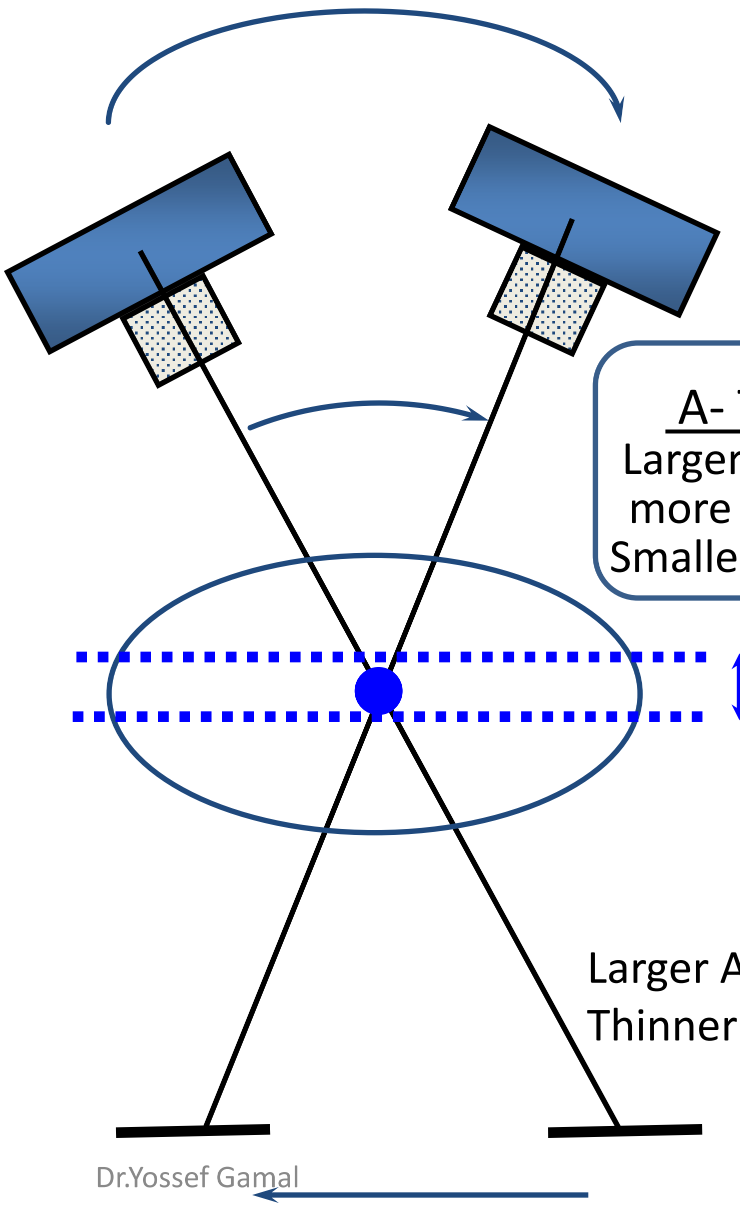


6) Height (level) of Cut

- Can be Adjusted by raising or lowering pivot

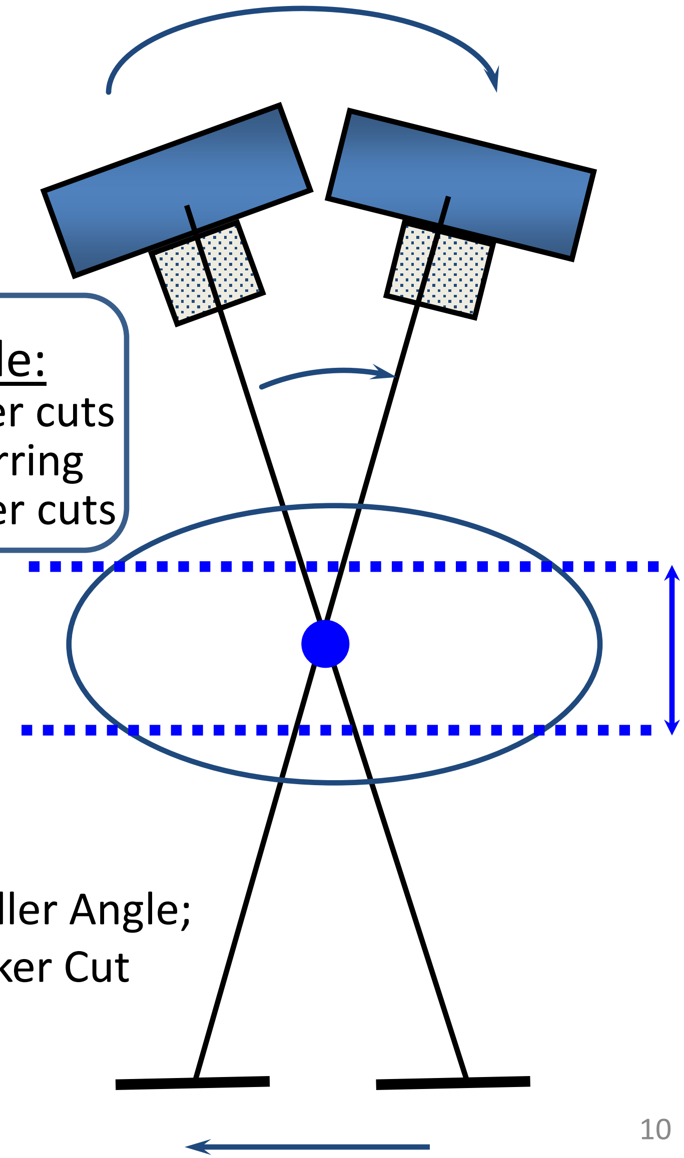


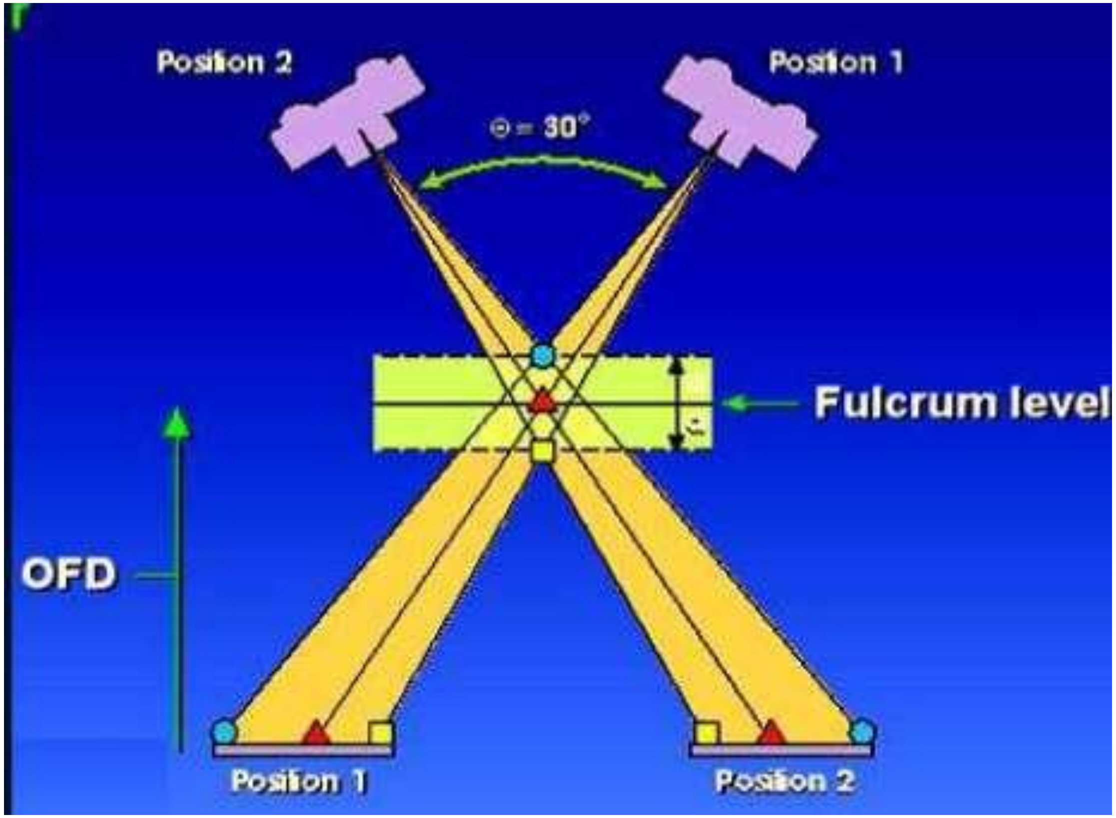
7)Factors affecting cut thickness:



A- Tomographic angle:

Larger angles yield thinner cuts
more motion = more blurring
Smaller angles yield thicker cuts





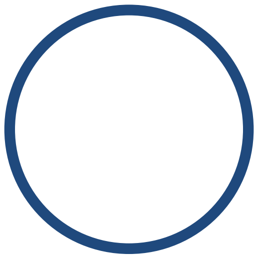
<u>B) OFD (fulcrum level):</u>
↑OFD (raising fulcrum) , will decrease cut thickness

<u>C) FFD :</u>
↓FFD , will decrease cut thickness

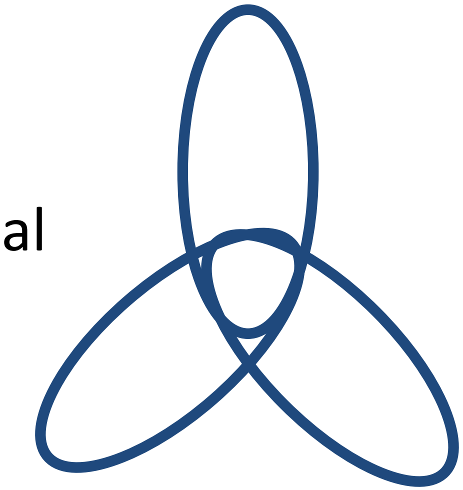
8) Complex Blurring

- Blurring improved by use of complex motions

circular



Hypocycloidal



tri-spiral

elliptical

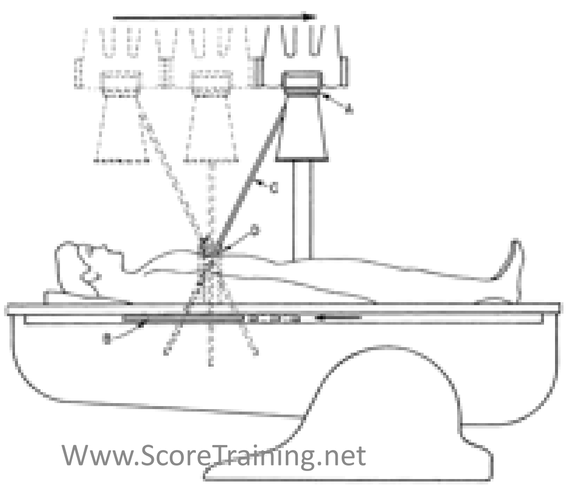
- The more complex the blurring motion
  - the sharper the tomographic image
    - better blurring
  - the more expensive the machine

9)Tomography contrast

- Film contrast in tomography is ↓
  – Due to spread of “off focus anatomy” over the film
  – So that↓Kv is used
  – Used in body part with high inherit contrast (e.g. inner ear)

10) Tomography patient dose

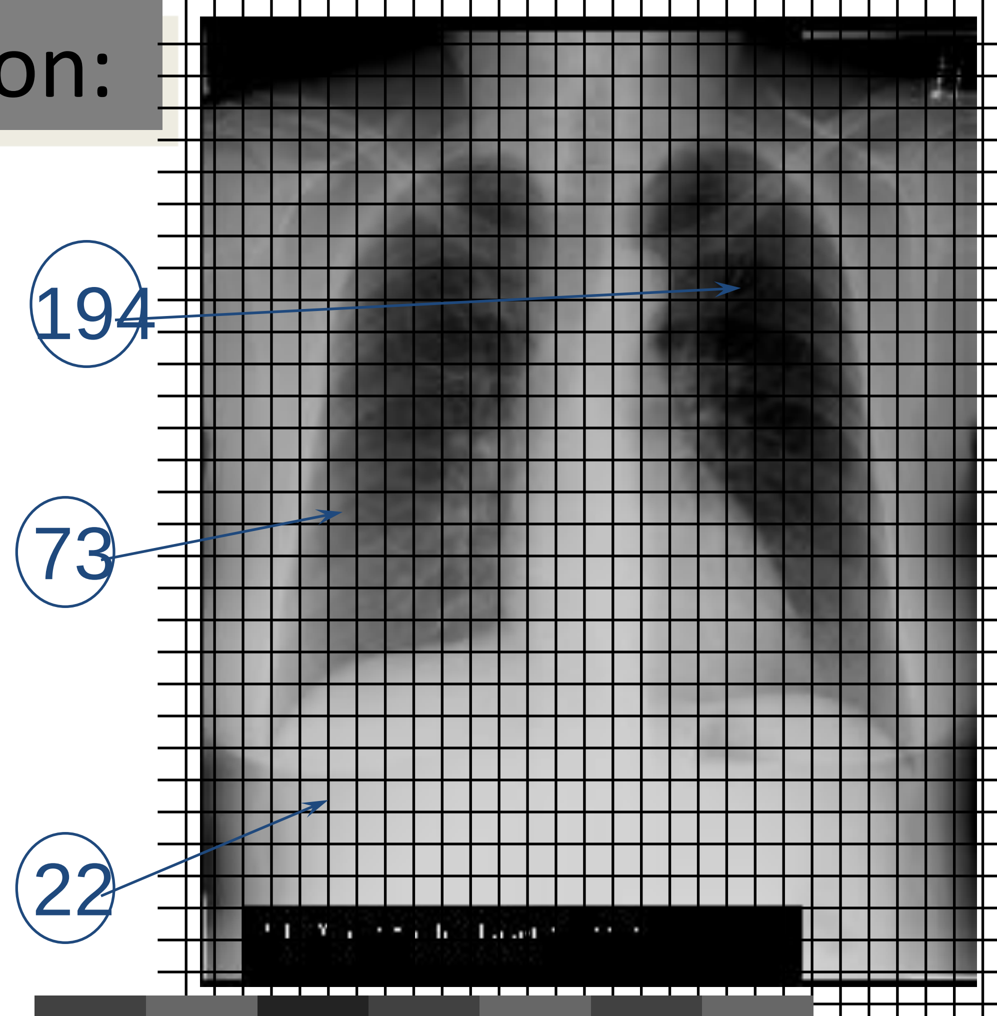
– During exposure , much of the beam is passing obliquely through the patient → patient’s dose is greater than conventional radiology



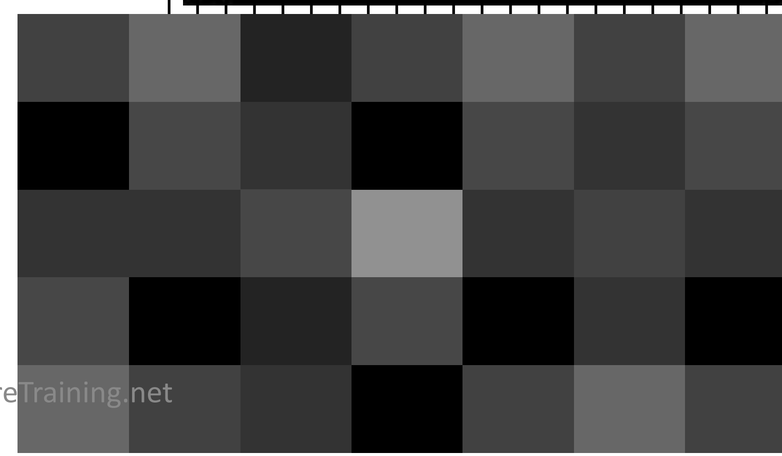
Digital radiography

Digital Image Formation:

- Digital image is divided into matrix of pixels
- each **pixel** is Assigned a value based on density (high values for dark pixels , and low values for bright pixels)
- The Value is the **average** of the pixel
  - Any spatial resolution within a pixel is lost
- Pixel values form the digital image

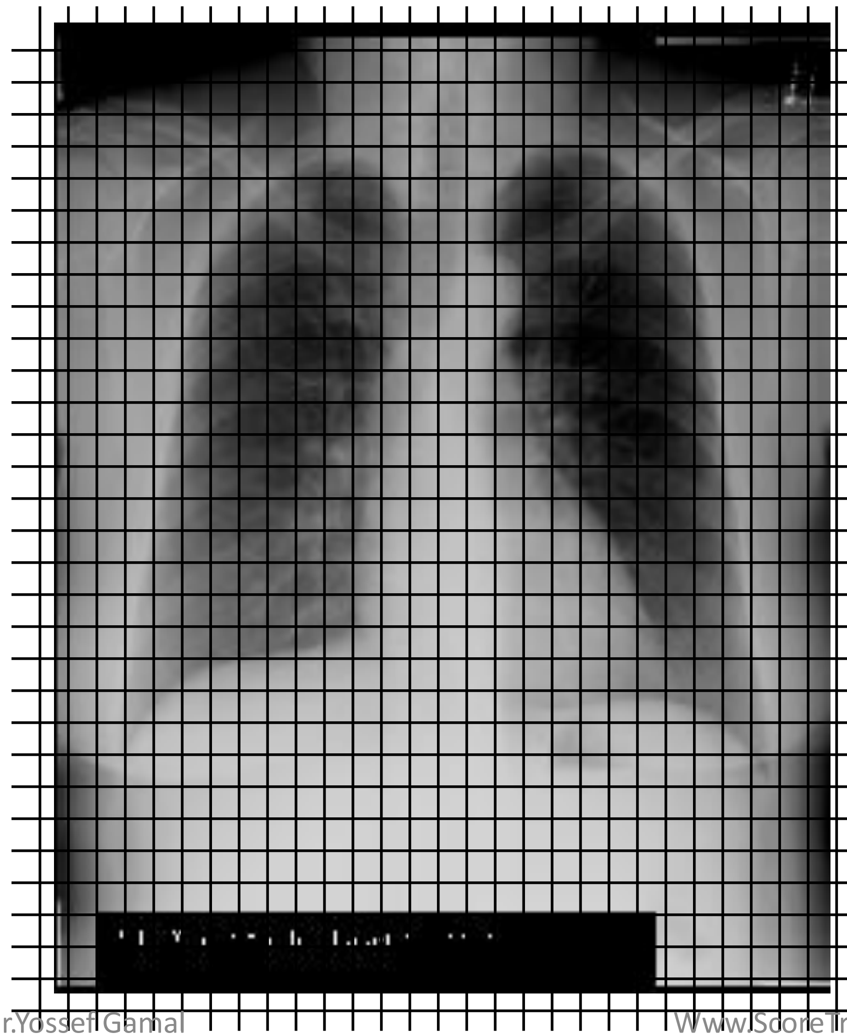


125	25	311	111	182	222	176
199	192	85	69	133	149	112
77	103	118	139	154	125	120
145	301	256	223	287	256	225
178	322	325	299	353	333	300

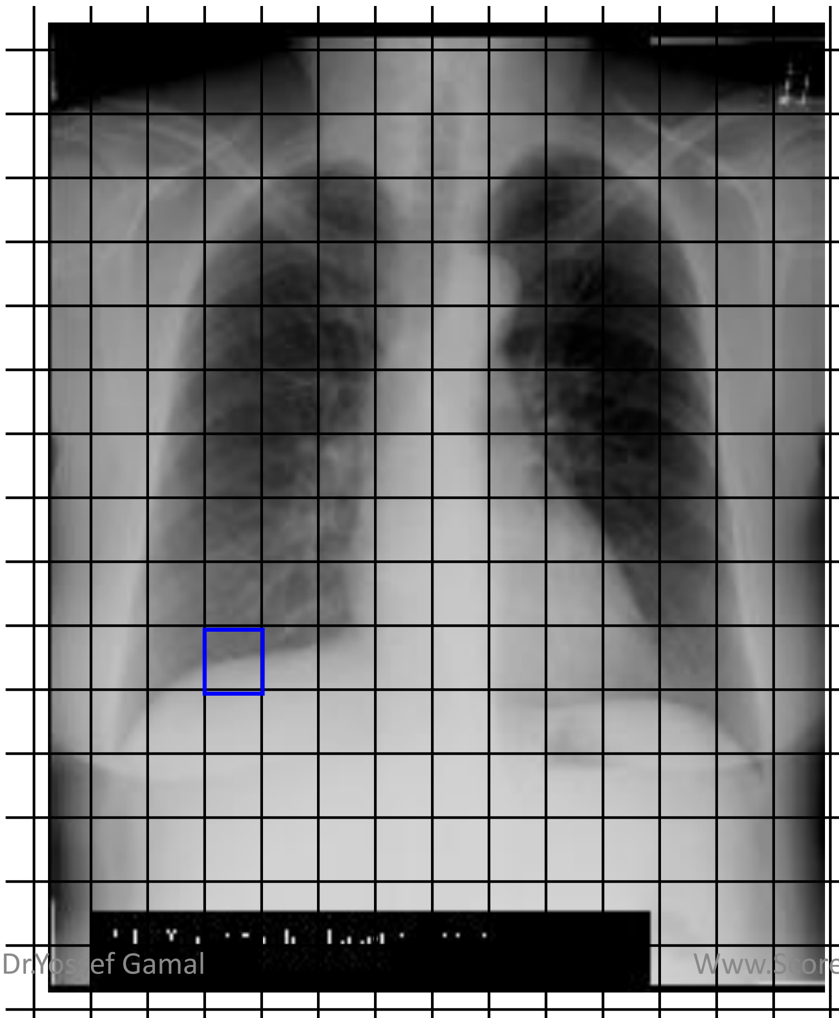


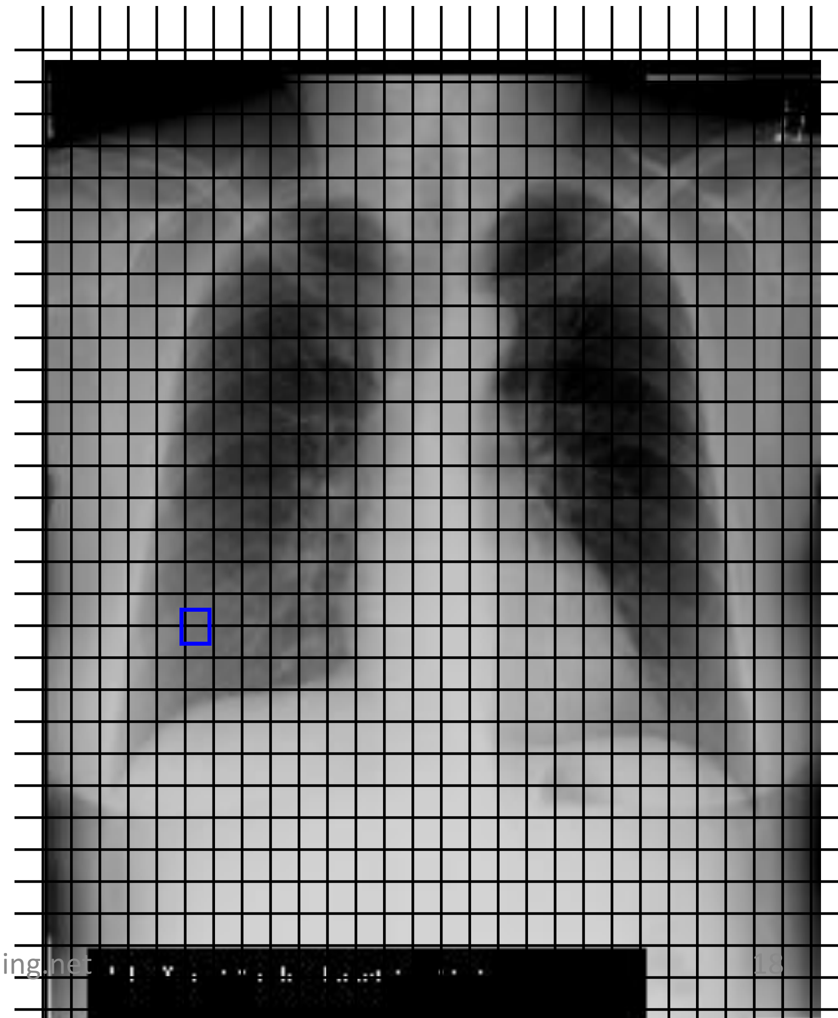
Digital Image Formation

digitalization ← Analogue Image



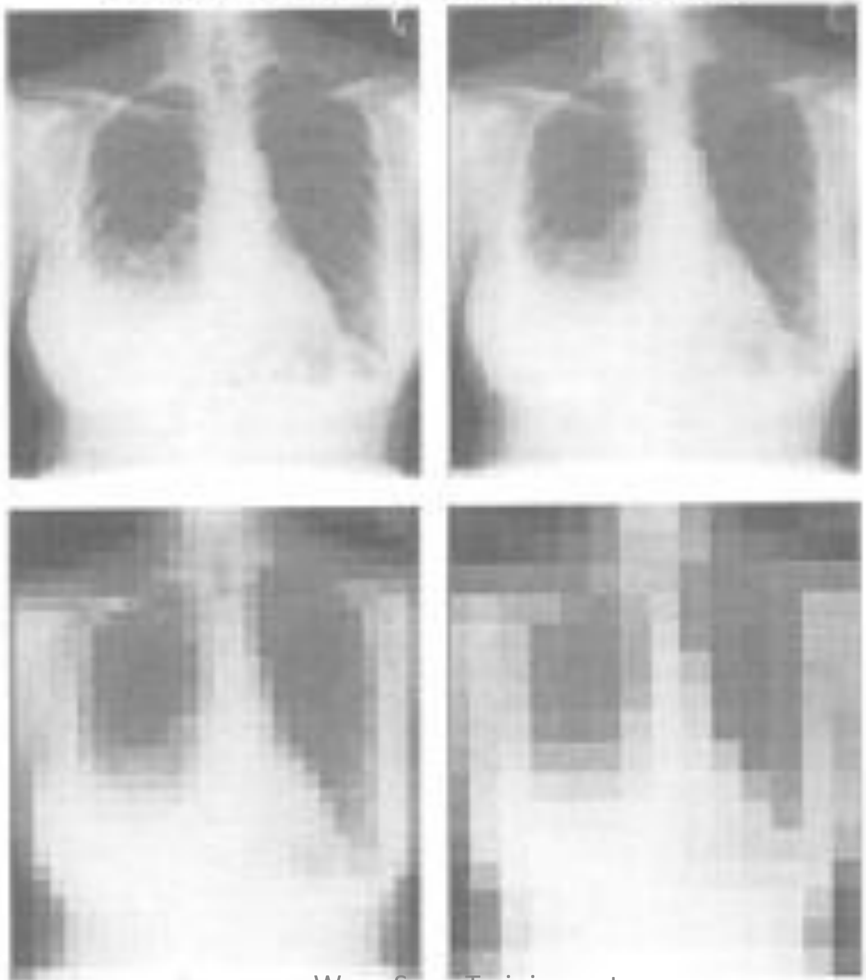
- Matrix size = number of pixels e.g. 512 x 512 matrix
- The more is the matrix size , the larger is the field of view (with fixed pixel size)
- The more is the matrix size , the smaller is the pixel (with fixed field of view)
- The finer the pixels , the more is the spatial resolution
- Objects smaller than the pixel size will not be seen





Effect of changing matrix size

1024^2, 64^2, 32^2, 16^2 matrices



Numerical and binary system

Numerical system is decimal (10 base numbers)

Binary system

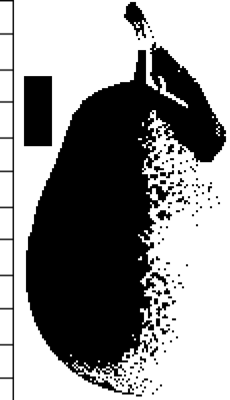
- **Fundamental method of computer storage**
- **two base numbers i.e. Only 2 allowable values**
  - 0
  - 1
- **Bit** = binary digit
  - Smallest binary unit
  - has value 0 or 1 only
- Computers do all operations with 0's & 1's
- no –ve power , no values between 0 & 1

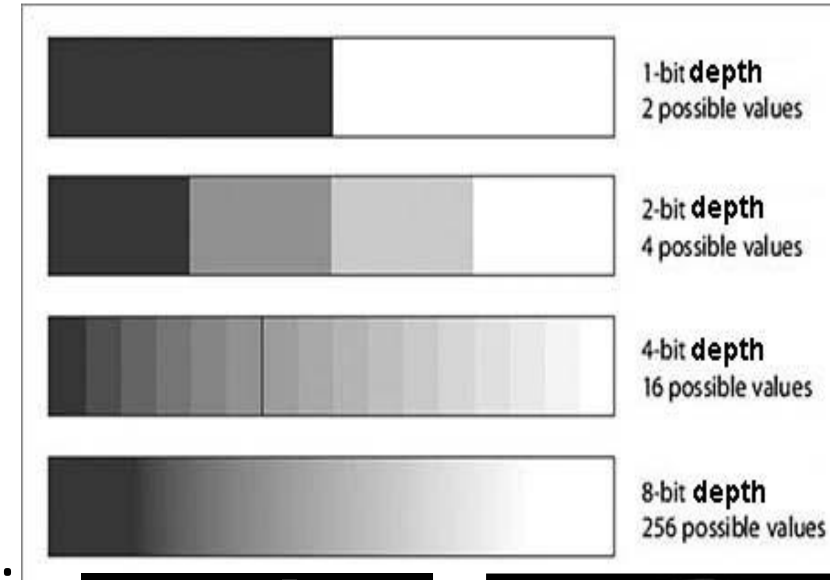
- **Bit depth** = bits per *pixels*
- – Single memory bit: each pixel has ONLY two possible values (0 or 1)
- – 4 bit depth: 16 possible values (2^4) , minimum = 0000 , maximum = 1111
- – 8 bit depth : 256 possible values (2^8) , i.e. 256 levels of gray could be displayed
- – The greater is the bit depth , the greater is the potential to display good contrast images
- – Factors affecting bit depth required:

1)Image noise : noisy images e.g. radionuclide, bit depth used is not greater than 8

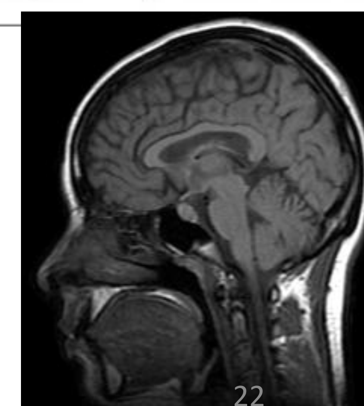
2)Dynamic range of the detectors (directly proportional)

0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
0	0	0	0	0	0	0	0	0	0	0	0	0	0	1	0
0	0	0	0	0	0	0	0	0	0	0	0	1	1	0	0
0	0	0	0	0	0	0	0	0	0	1	1	0	0	0	0
0	0	0	0	0	0	0	0	1	1	0	0	0	0	0	0
0	0	0	0	0	0	1	1	0	0	0	0	0	0	0	0
0	0	0	0	0	1	0	0	0	0	0	0	0	0	0	0
0	0	0	1	1	0	0	0	0	0	0	0	0	0	0	0
0	1	1	0	0	0	0	0	0	0	0	0	0	0	0	0
0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0

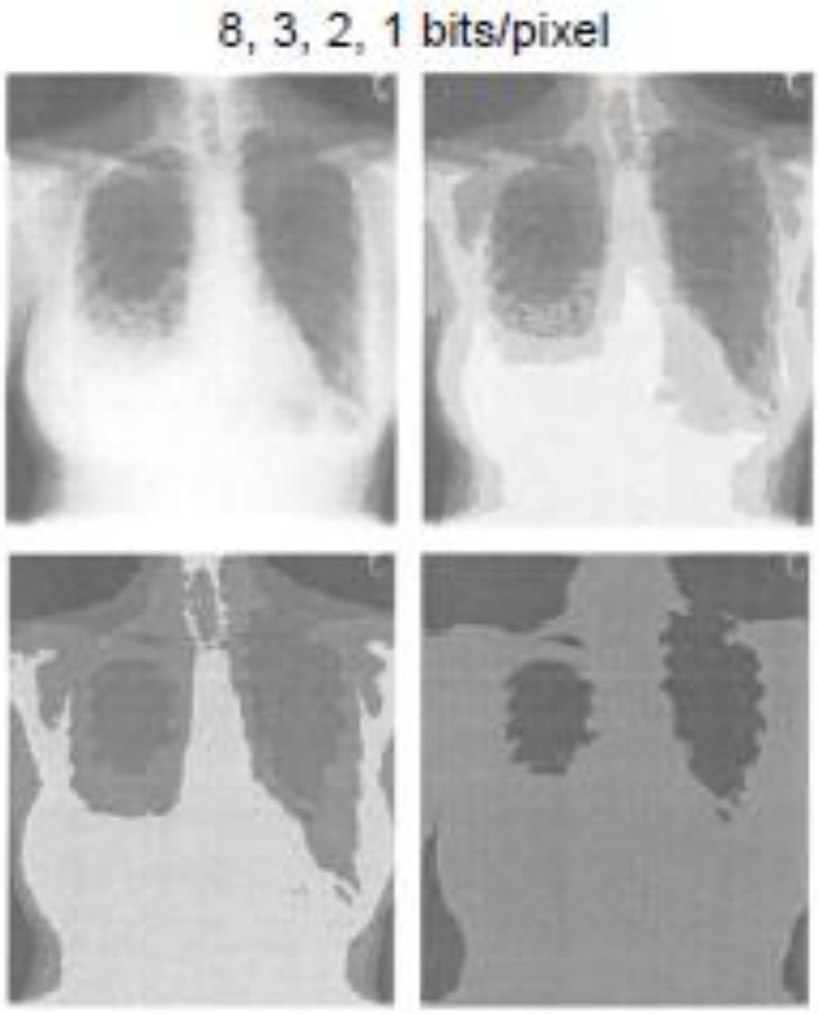





4 grade shades


256 grade shades

Effect of changing bit depth

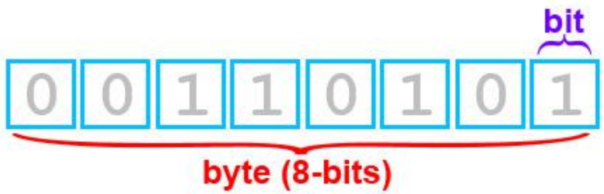


- **Byte** : method of expression of computer memory
  - 1 bytes = 8 bits
- Example:

A CT image with matrix size of 512 x 512 pixels , and bit depth of 12/pixels , how much is the memory required to store this image?

$$= \frac{\text{total number of bits saved}}{8} = \frac{512 \times 512 \times 12}{8}$$

= 393216 bytes = 393 KB



N.B: IN 8 bit depth system , memory required to store each pixel = ...

Analog vs. Digital Images

- Analog
  - continuous gray shade information

- Digital
  - Discrete gray shade information

Compression

- Benefits:

1)Reduce storage requirements

2) Faster transmission times

- Types:

<u>1)Lossless (reversible):</u>

-Image can be restored into identical version of original

-Allows reduction of size up to 2-3 times (depending on the image quality)

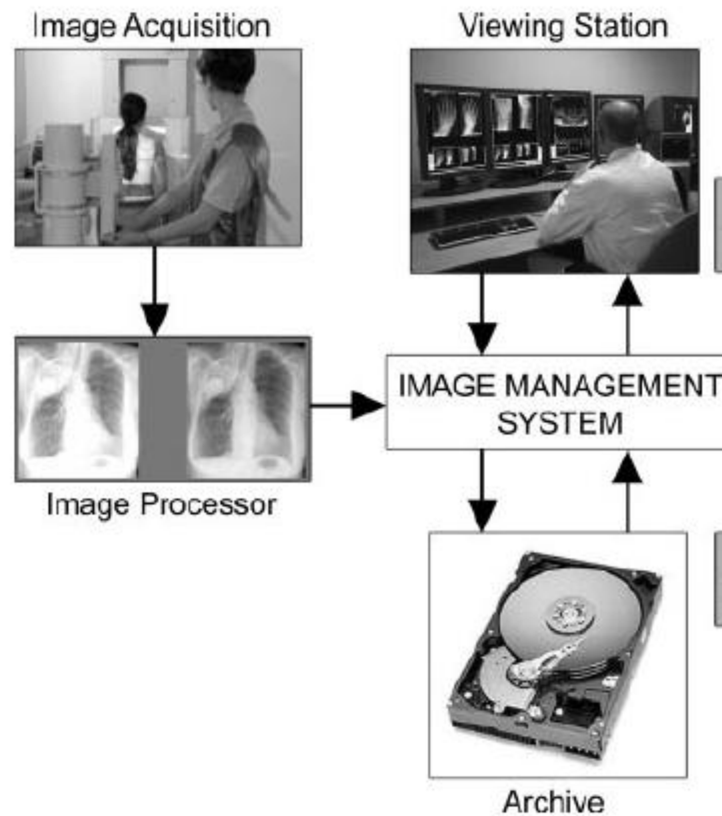
<u>2)Lossy (irreversible):</u>

-Displayed image does not perfectly reproduce the original

- Allow reduction of size up to 40 times

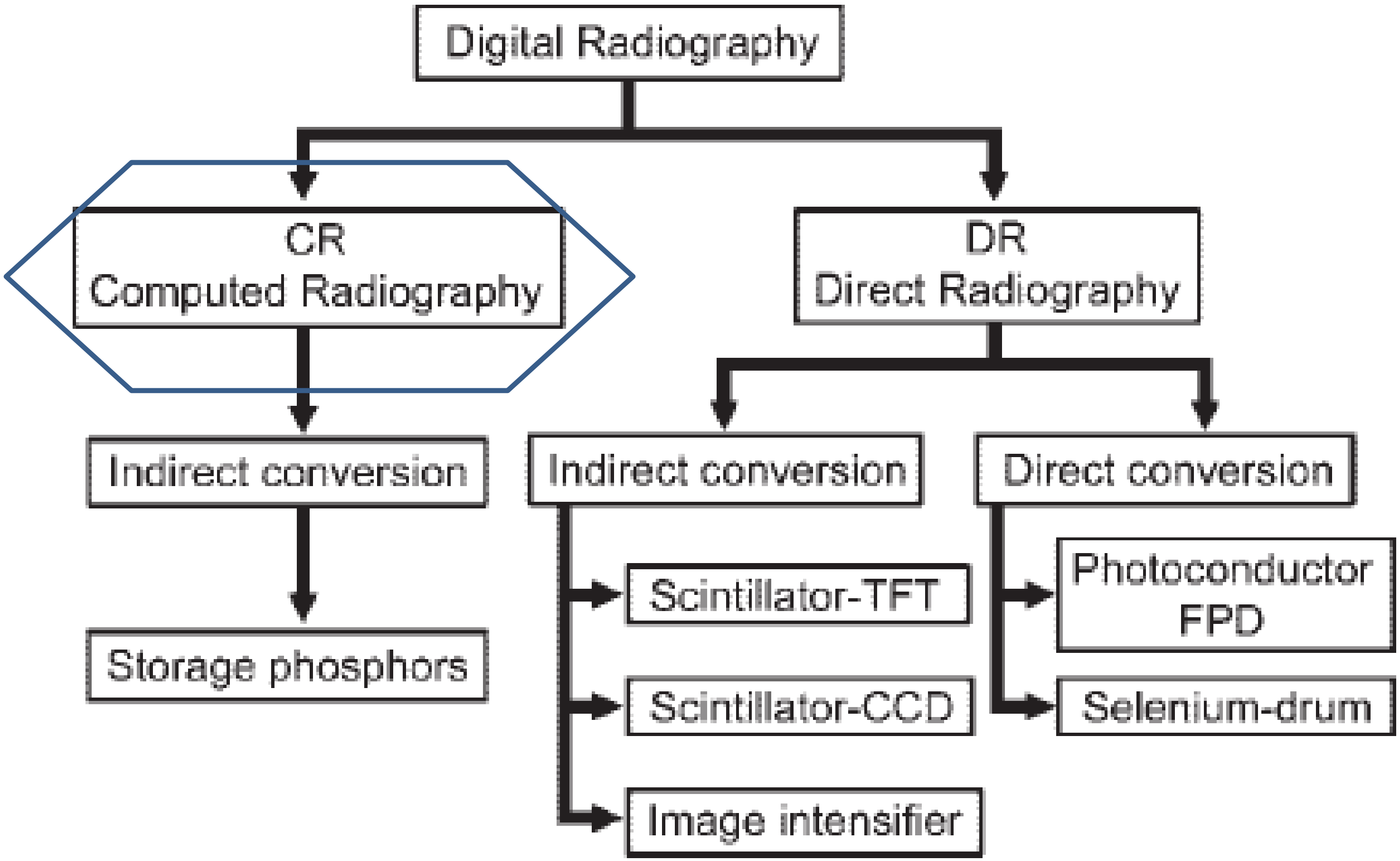
-May not be acceptable to be used in primary diagnosis

Principles of Digital Radiography



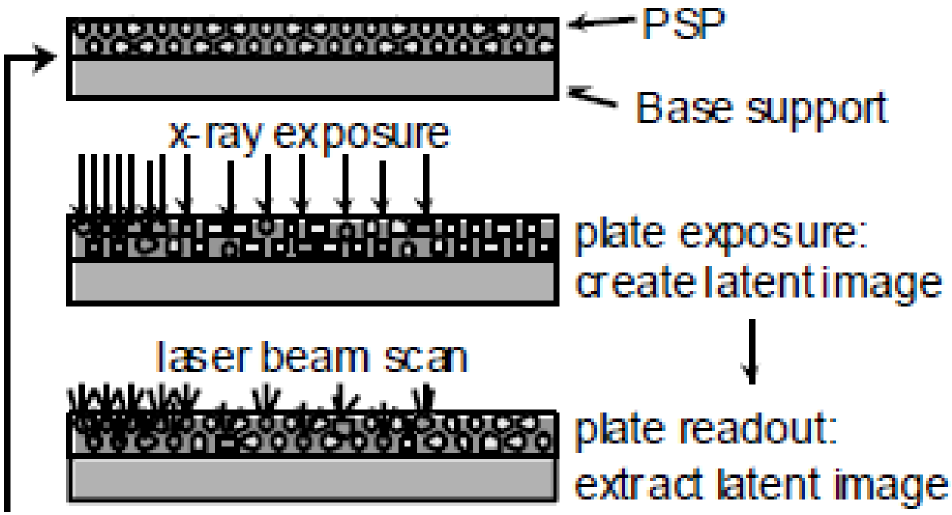
- Digital imaging comprises four steps:
- Generation: energy absorbed by the digital detector is transformed into electrical charges → recorded and digitized
- Processing: organize raw data into a meaningful image.
- Archiving: images are sent to a digitized storage archive
- Presentation: digitally on a workstation (or hard copy), Image can be manipulated e.g. zooming, windowing etc.

Types of digital radiography



Computed radiography (CR)

- Depends on phosphor imaging plates which:
  - ✓Replace film & screen in the cassettes
  - ✓Re-usable
  - ✓Uses conventional bucky & x-ray equipment



Computed radiography (CR)

Detectors: Storage phosphor image plates:

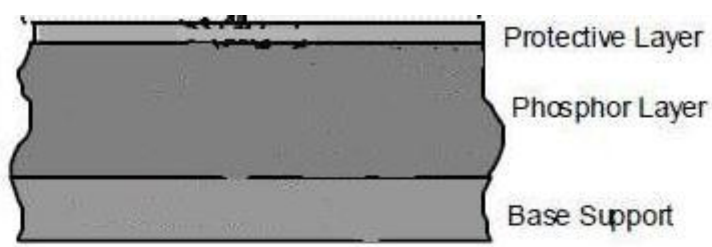
-crystal of barium flourohalide doped in europium

-halide is 85% bromide , 15% iodide (photostimulable phosphors)

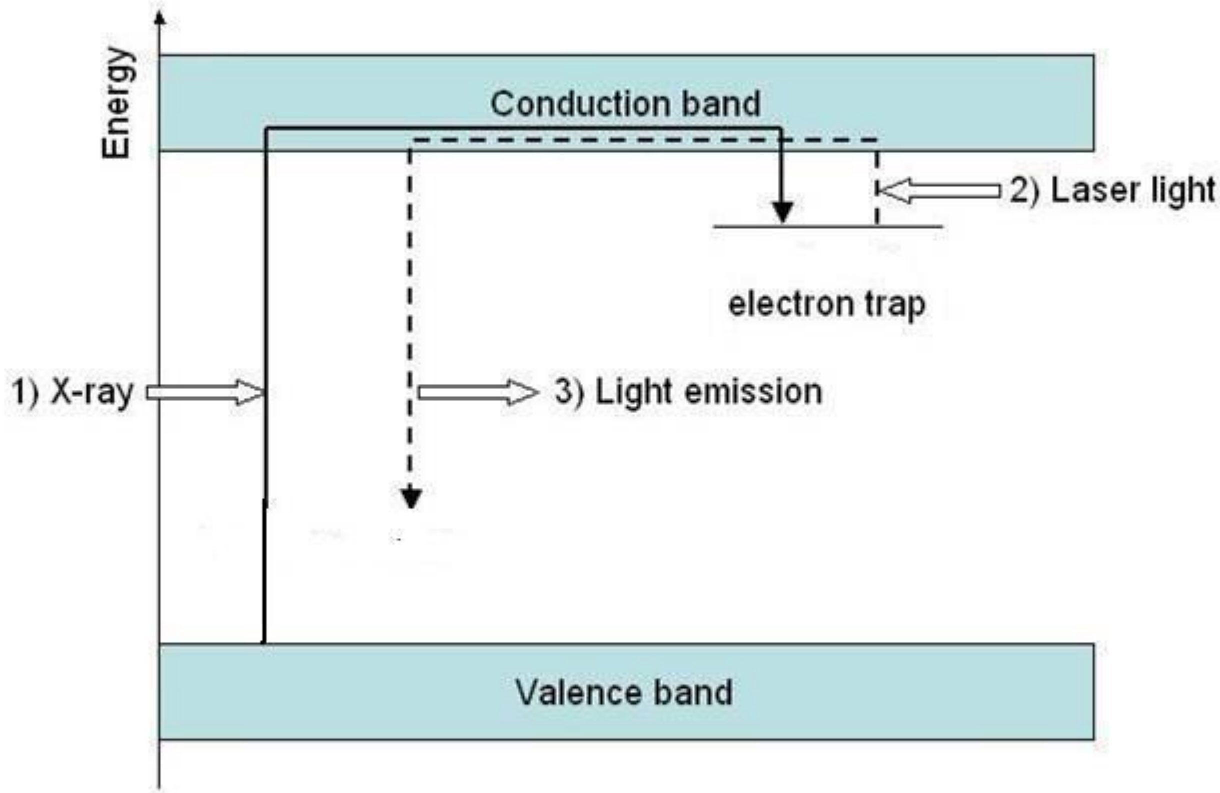
-present in unstructured way

-laid down in a base (0.3 mm thickness)

-a surface coat is present to protect from damage

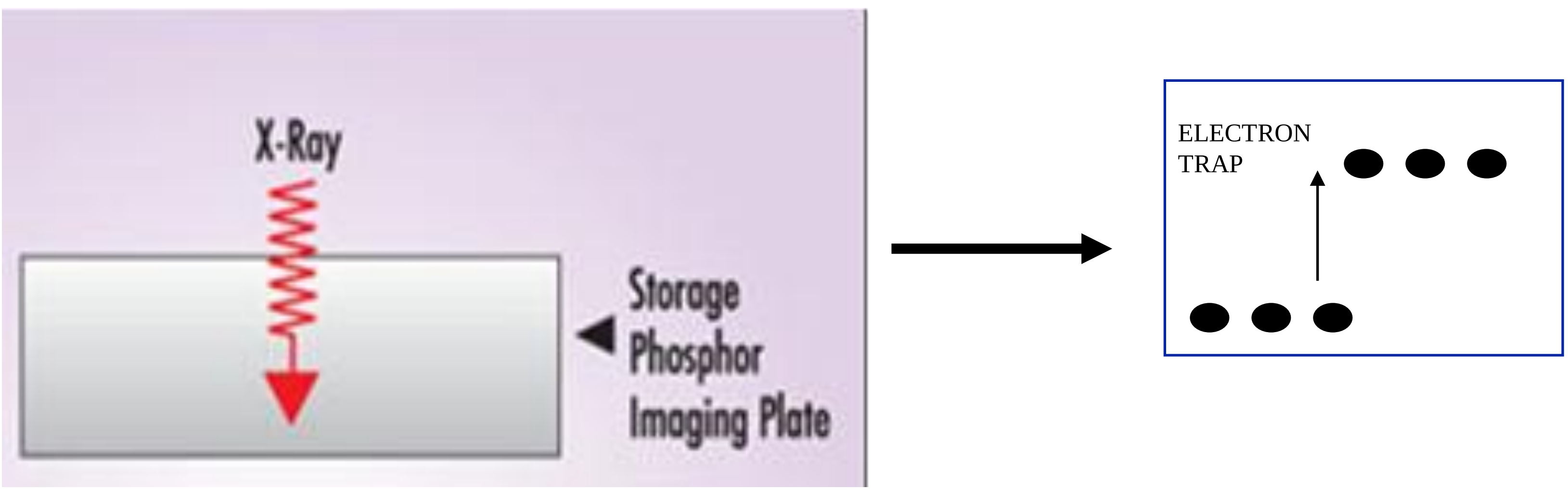


photostimulable phosphor



Exposure process :

- X-ray absorbed and temporarily stored by bringing electrons to a higher energy levels (electrons traps)



<u>Readout process</u>

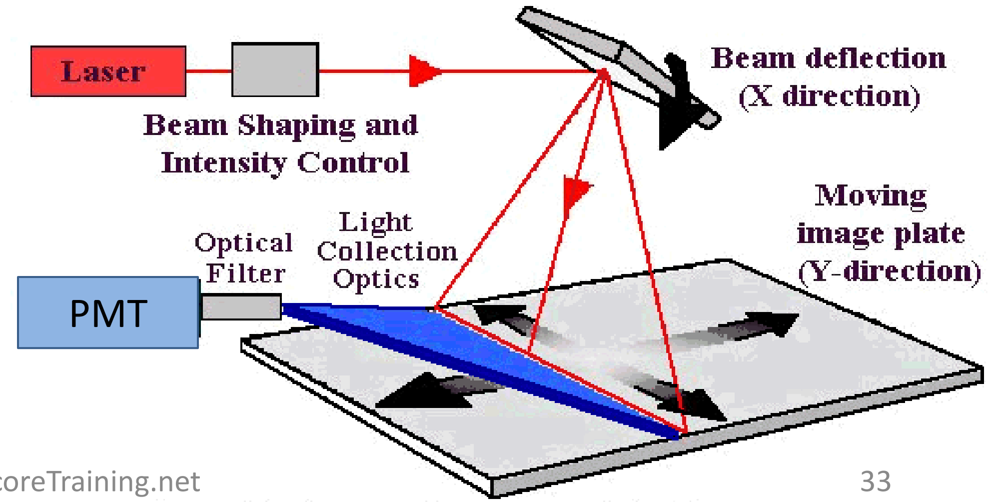
•Done By using flying spot scanner

•Separate step

•Detective layer is scanned pixel by pixel using high energy laser beam of specific wave length (using rotating mirror)

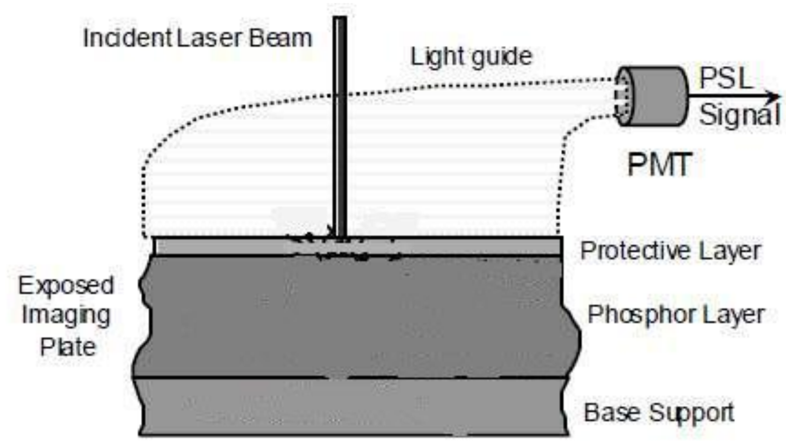
•Stored energy is set free as emitted light

•Intensity of emitted light indicates amount of radiation incident on phosphor at each location

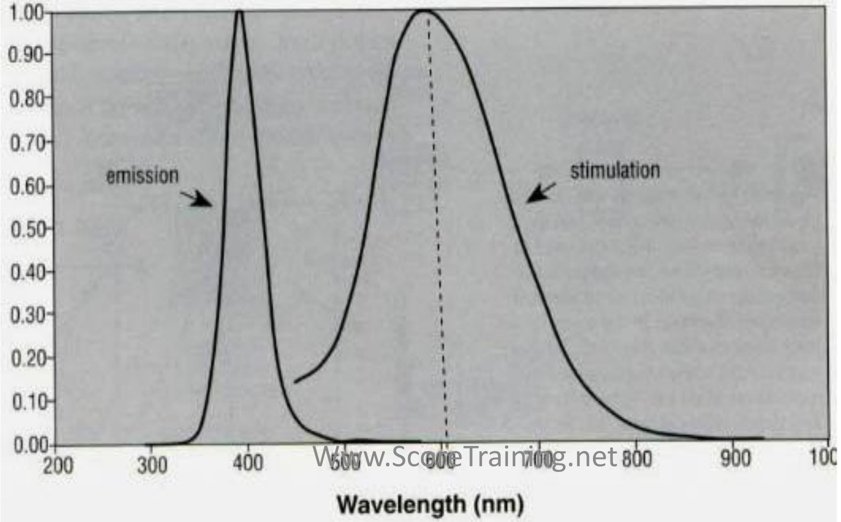


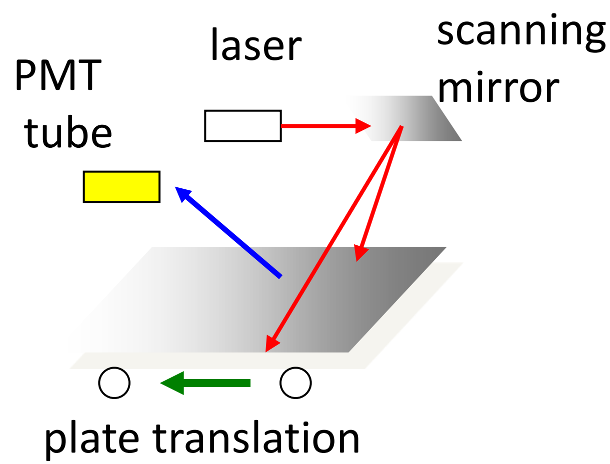
Schematic diagram of the components in a flying-spot scanner.

- Light guide (optical fibers) convey the light to array of photomultiplier tubes which collect the light and measure its intensity and then convert it into electrical charge

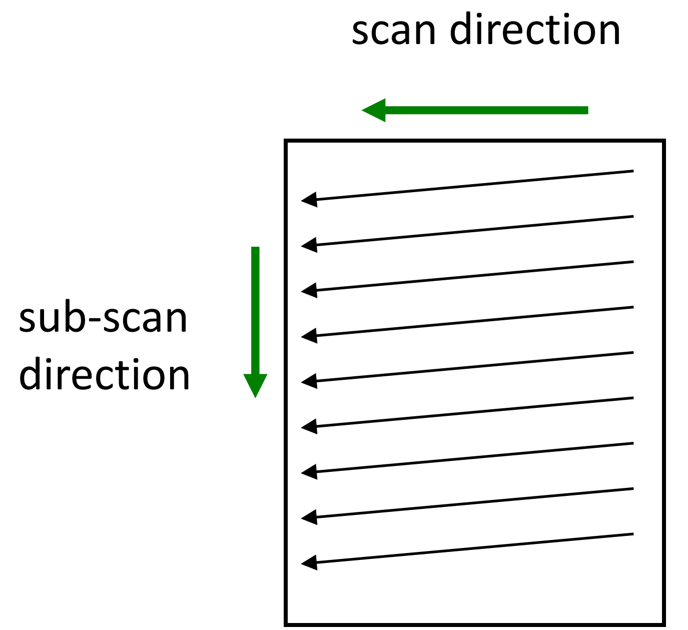


- The emitted light should have wave length different from that of the laser beam so that Only color of light emitted by phosphor measured by PMT
- Most phosphors emits blue light → needs laser emitting red light





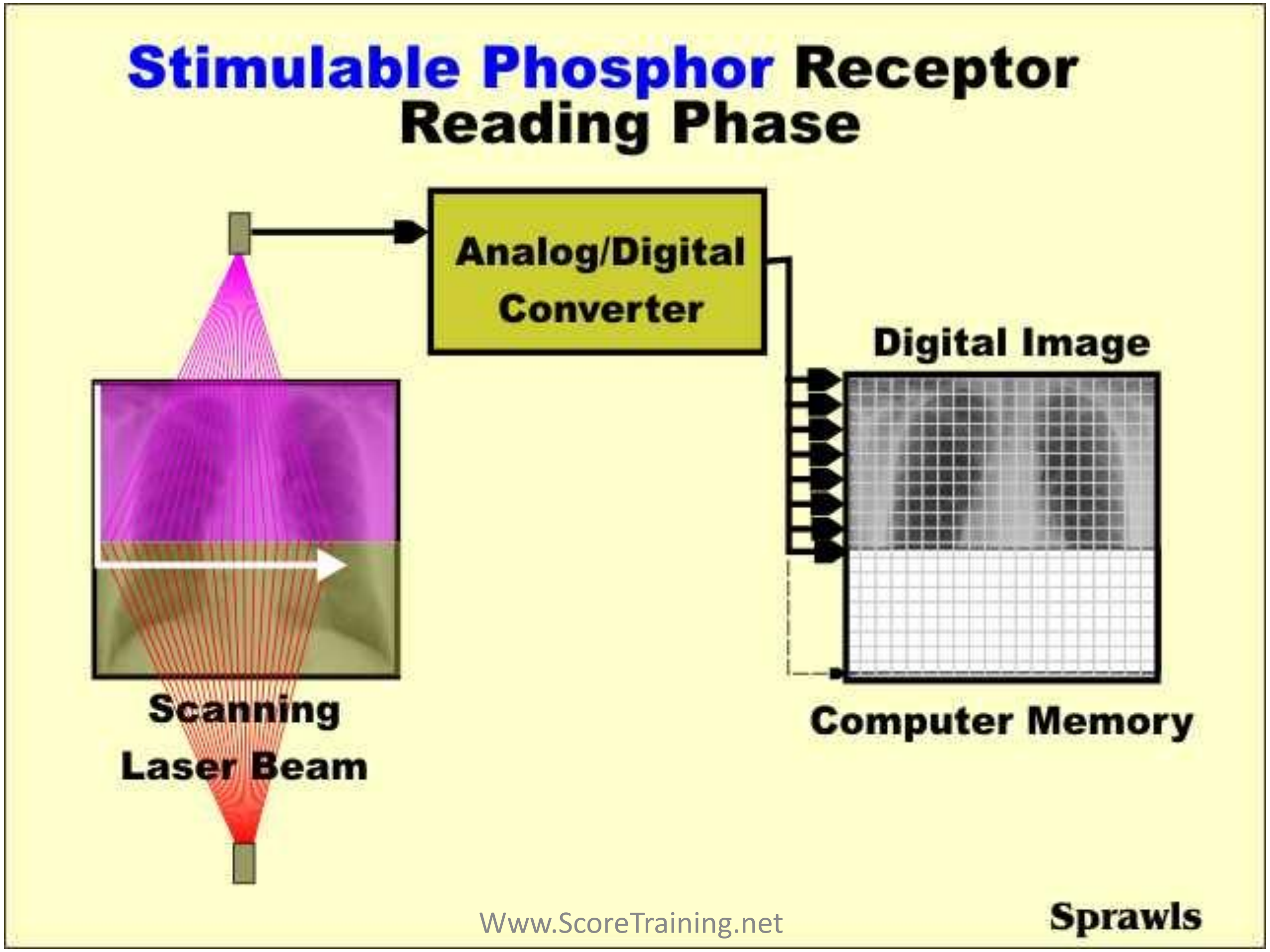
Read



CR plate

Position of light emitting center is determined from the time at which the light is received

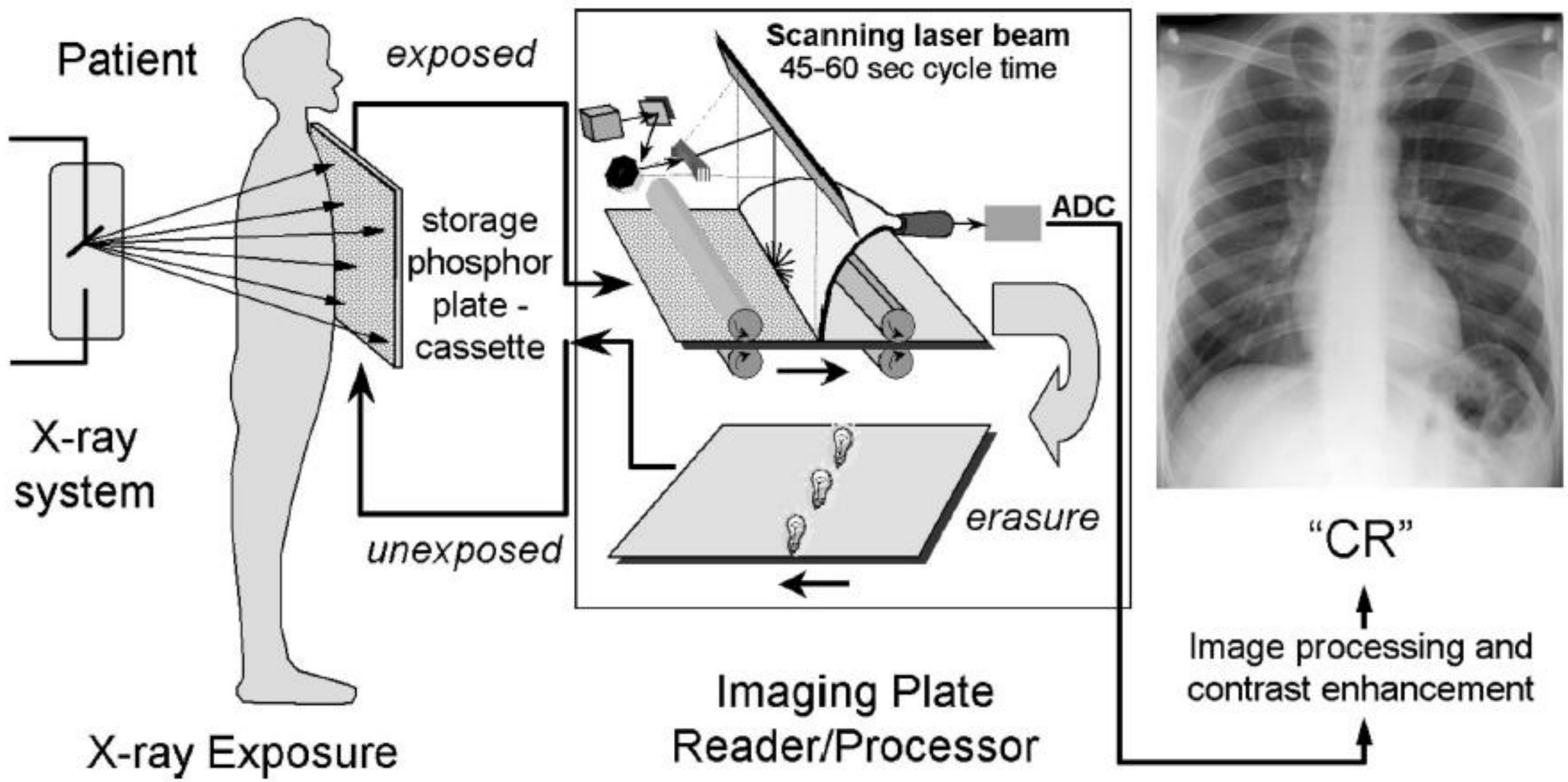
Then The charge output of PMT is digitalized by analogue to digital convertor



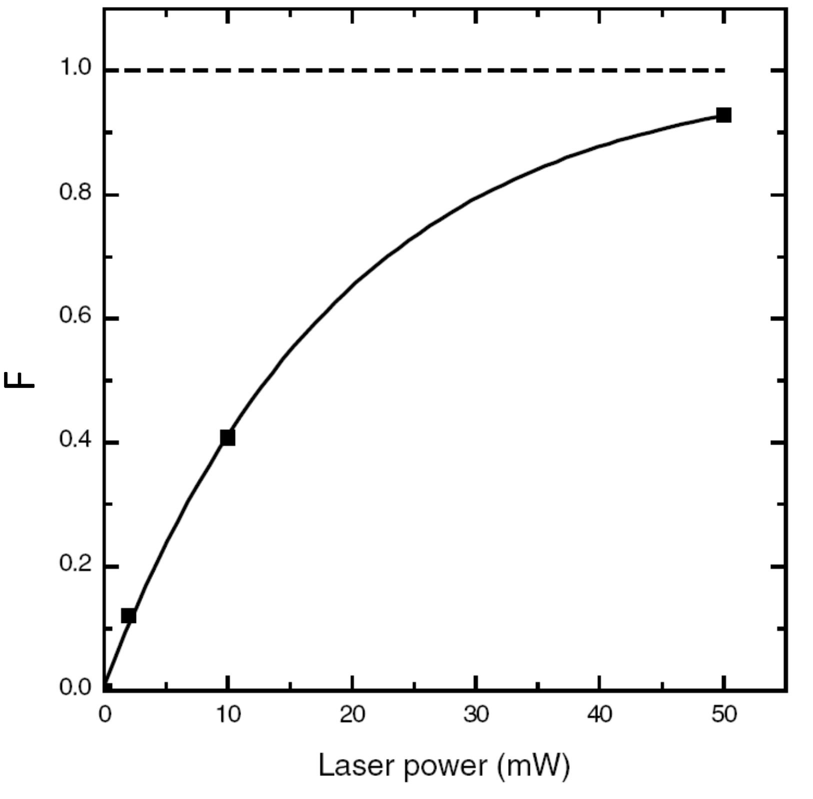
- <u>Following the read cycle , the residual plate signal is erased by exposing it to bright light (why?)</u>

Erase

Summary

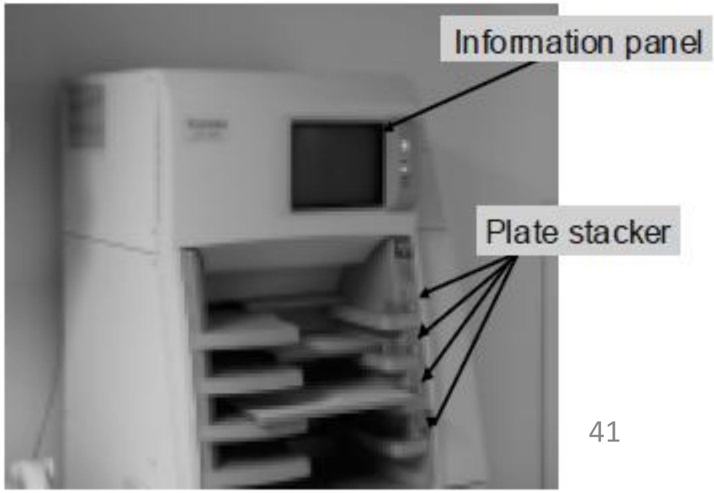

Patient
exposed
X-ray system
storage phosphor plate - cassette
unexposed
X-ray Exposure
Scanning laser beam
45-60 sec cycle time
ADC
erasure
Imaging Plate Reader/Processor
"CR"
Image processing and contrast enhancement

- Discharge fraction *F*
  - Fraction of trapped energy released by stimulating laser light
  - Depends directly on laser power incident on the plate (not linear)
  - *F* depends also on
    - time to read a plate
    - imaging plate type
    - the plate size
  - Typical figure for *F* is 0.5



- Notes :

- The energy stored in the detective stays for several hours depending on the physical properties of the phosphor crystals, yet , readout process should be done immediately before the stored energy decrease over time
- Readout process of a 14“ x 17“ image plate takes 30-40 seconds (maximum workload of 90-120 image plate/h)
- Stacking reader are available in which several cassettes is placed in queue for automatic feed in the reader

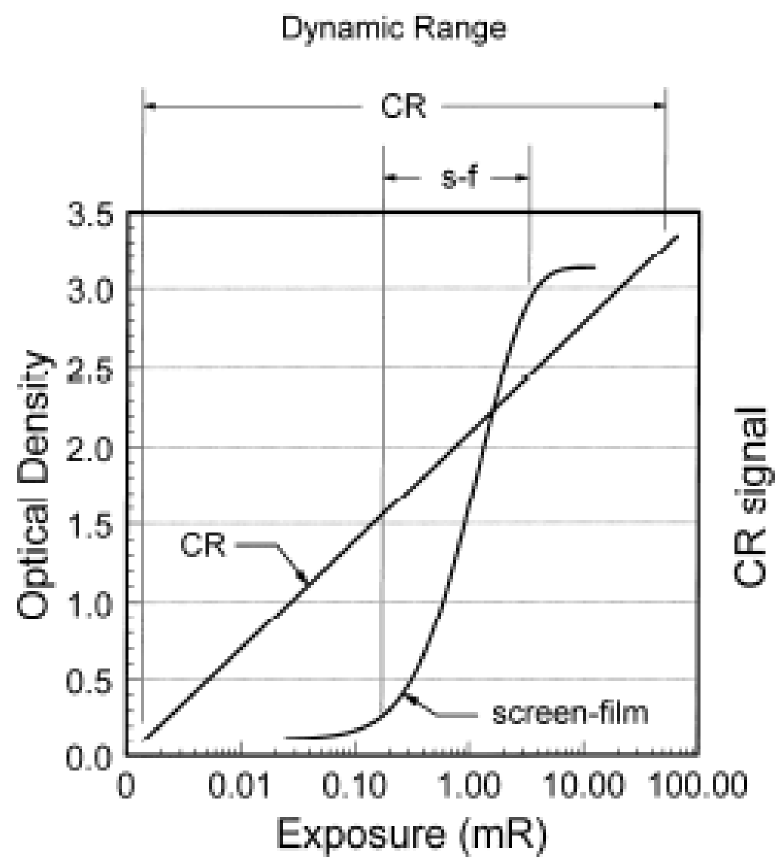


CR image processing

- Photostimulable phosphors have very wide latitude (dynamic range) = 10000:1

i.e. able to record photon intensities varying by factor of 10000:1 , with linear relationship between dose and response (In film screen = 16:1)

i.e. very wide latitude→ very low film γ (0.4)



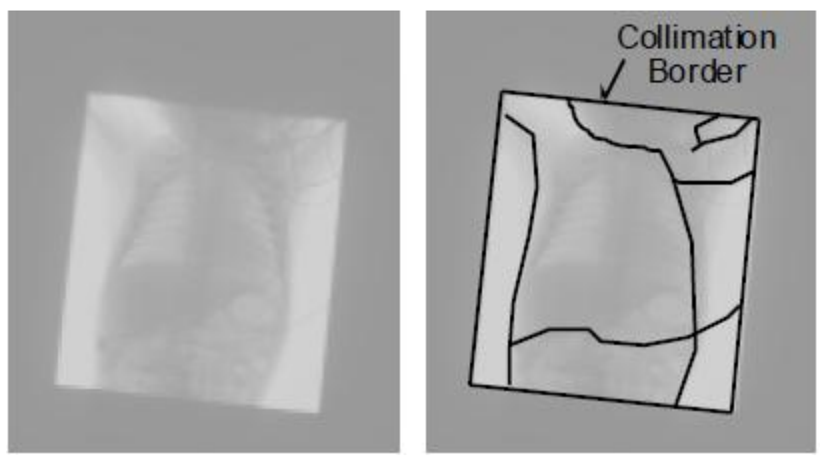
Steps of image processing

<u>1- image segmentation:</u>

Detect the edges of intensities of x-ray beam

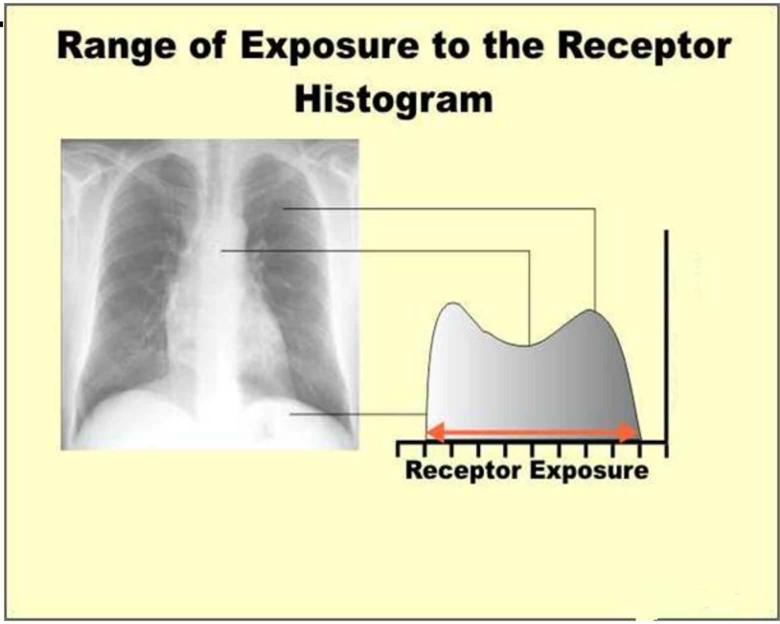
Outside that edges → signal is ignored

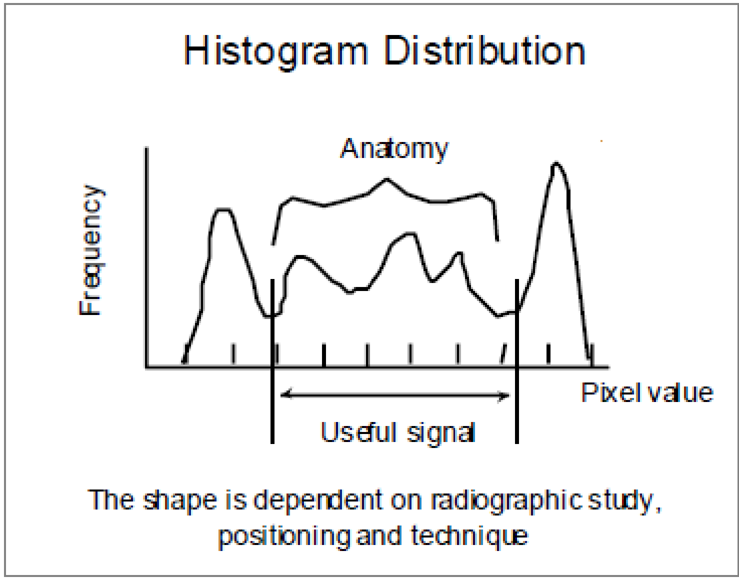
i.e. Separate raw radiation from anatomy



2) Histogram analysis of intensities within the collimated area

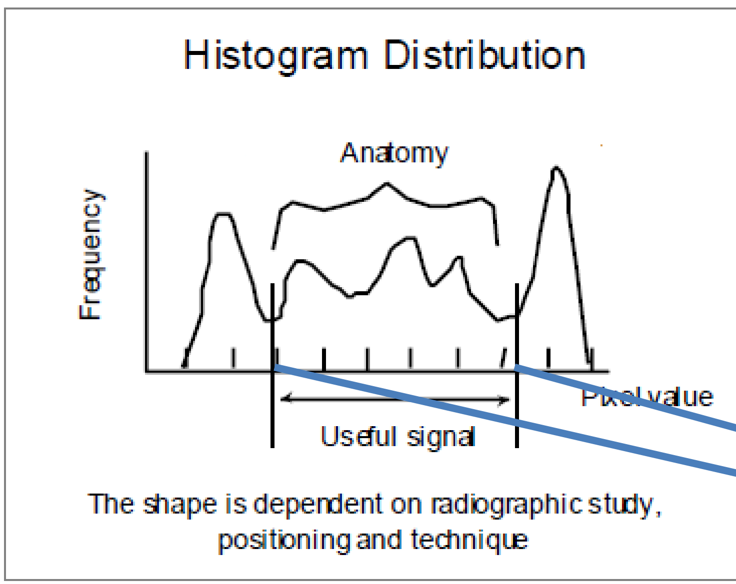
- Plot of gray scale value vs. the frequency of occurrence within the defined area in the image
- Shape of the histogram is anatomy specific
- Sets minimum and maximum "useful" pixel values
- Very ↑signal (outside the body) and very↓ signal (e.g metal) are rejected

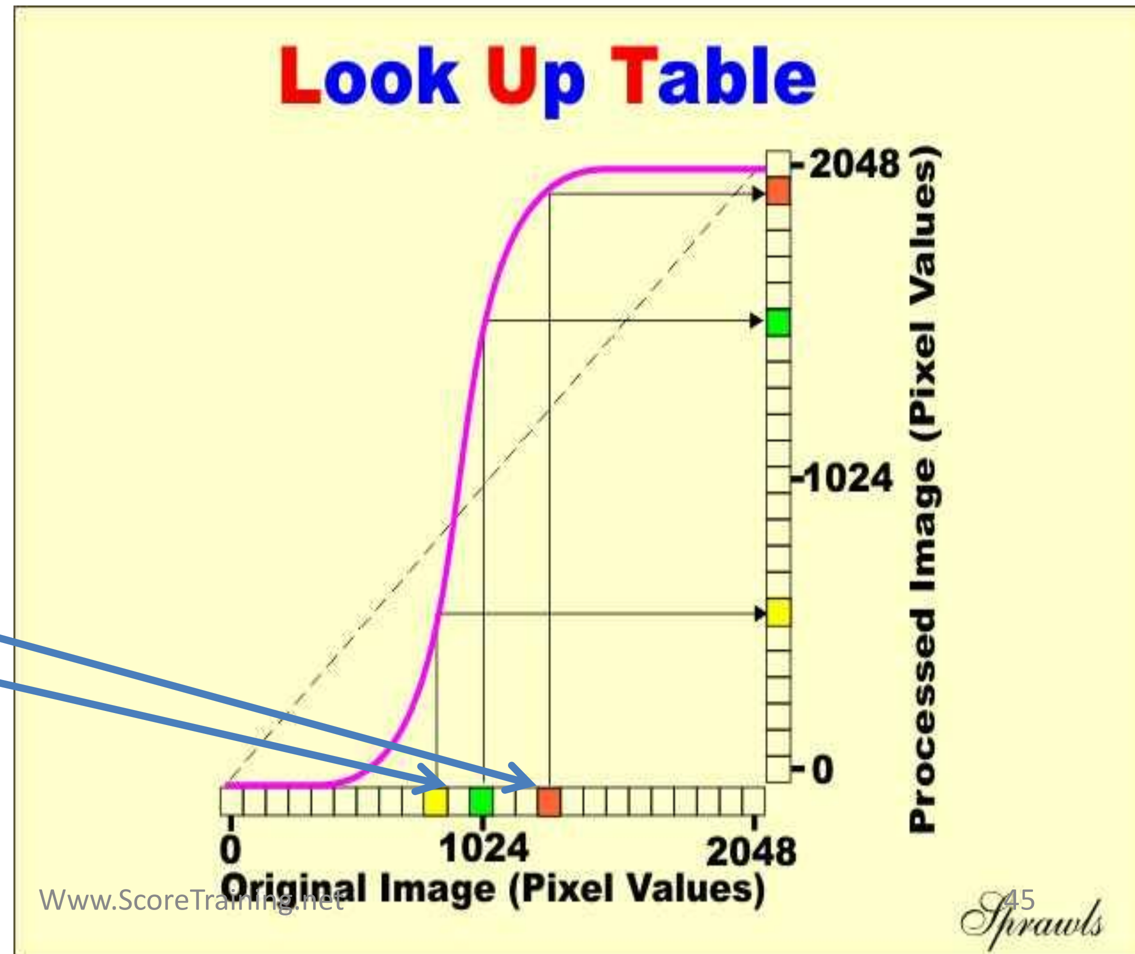




3) Look up table transformation (Tone Scaling):

- Map useful intensity values to the linear portion of the characteristic curve
- This process is optimized for each particular projection
- Determine the contrast in CR

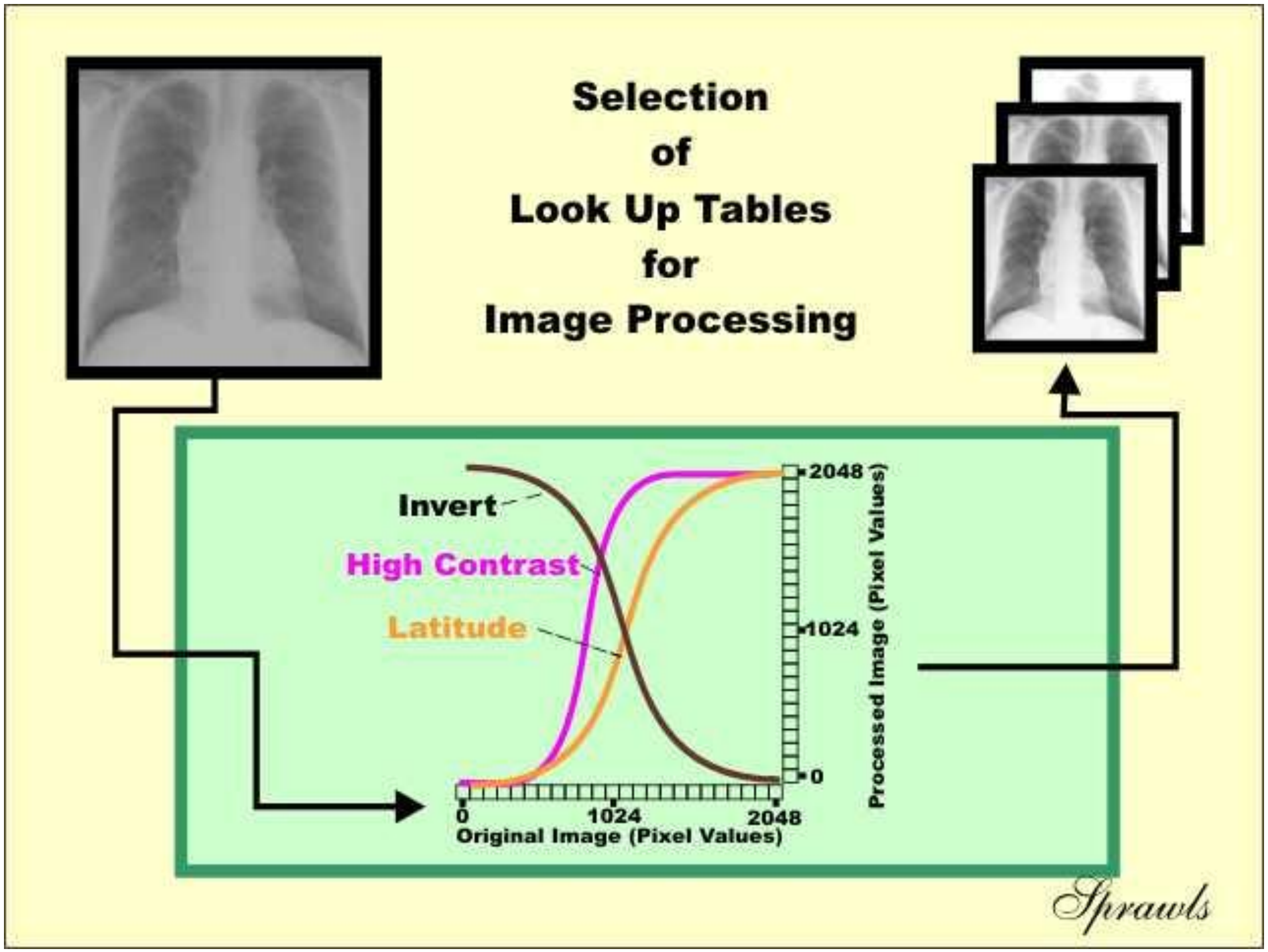


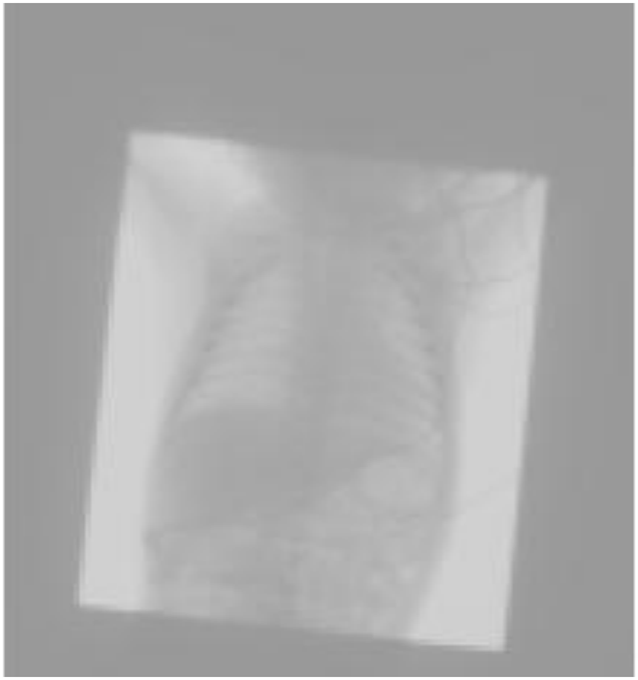


LUT Selected depends on

- Body part
- projection

User can also alter LUT manually





PREPROCESSING

POSTPROCESSING

CR spatial resolution

- Spatial resolution of CR is less than conventional film screen radiology due to Scattering of laser light in the phosphor layer
- To increase CR resolution (e.g. mammo):

1- use thinner phosphor

2- decrease diameter of scanning laser beam

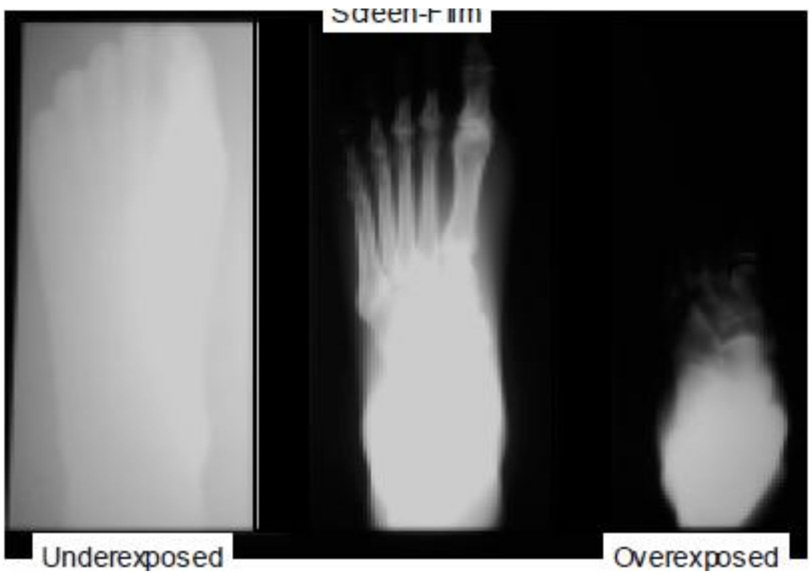
In radiographs with large image plates (e.g. chest & abdomen) low resolution plates are used to decrease imaging time and file size

CR screen sizes, Image size, and Pixel, and study size, following 12-bit logarithmic quantization.

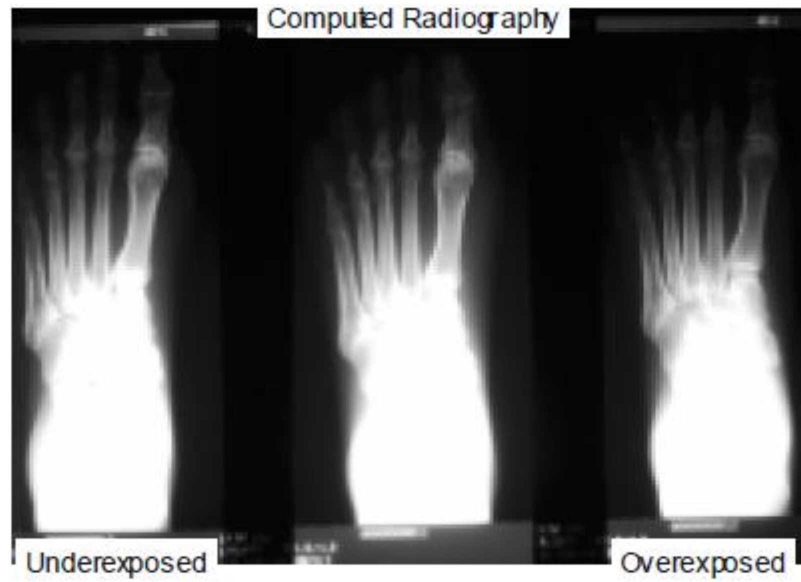
Screen Size	Image Size	Pixels/mm	Study Size
35 cm x 43 cm	2048 x 2500	5.8	11.5 MB
24 cm x 30 cm	2048 x 2500	8.3	11.5 MB
18 cm x 24 cm	1792 x 2392	9.9	9.7 MB

Advantages of CR:

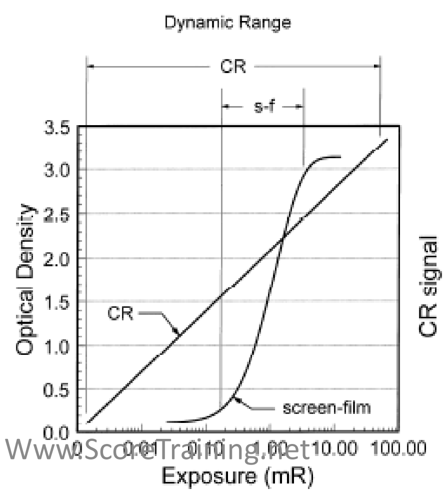
1- Wide dynamic range (latitude) → decrease rates of failed x-ray exposures



Film screen radiography

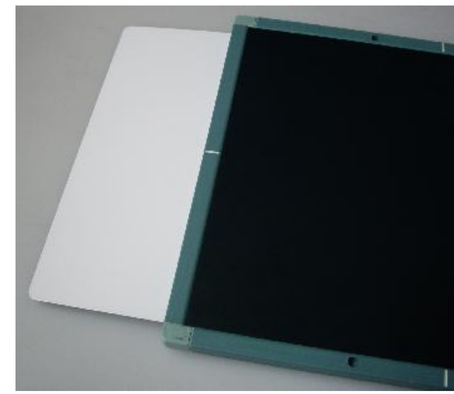


CR



2-Cassette based , so that:

- Easily integrated into existing radiographic devices
- Highly mobile (used in bed side exams)



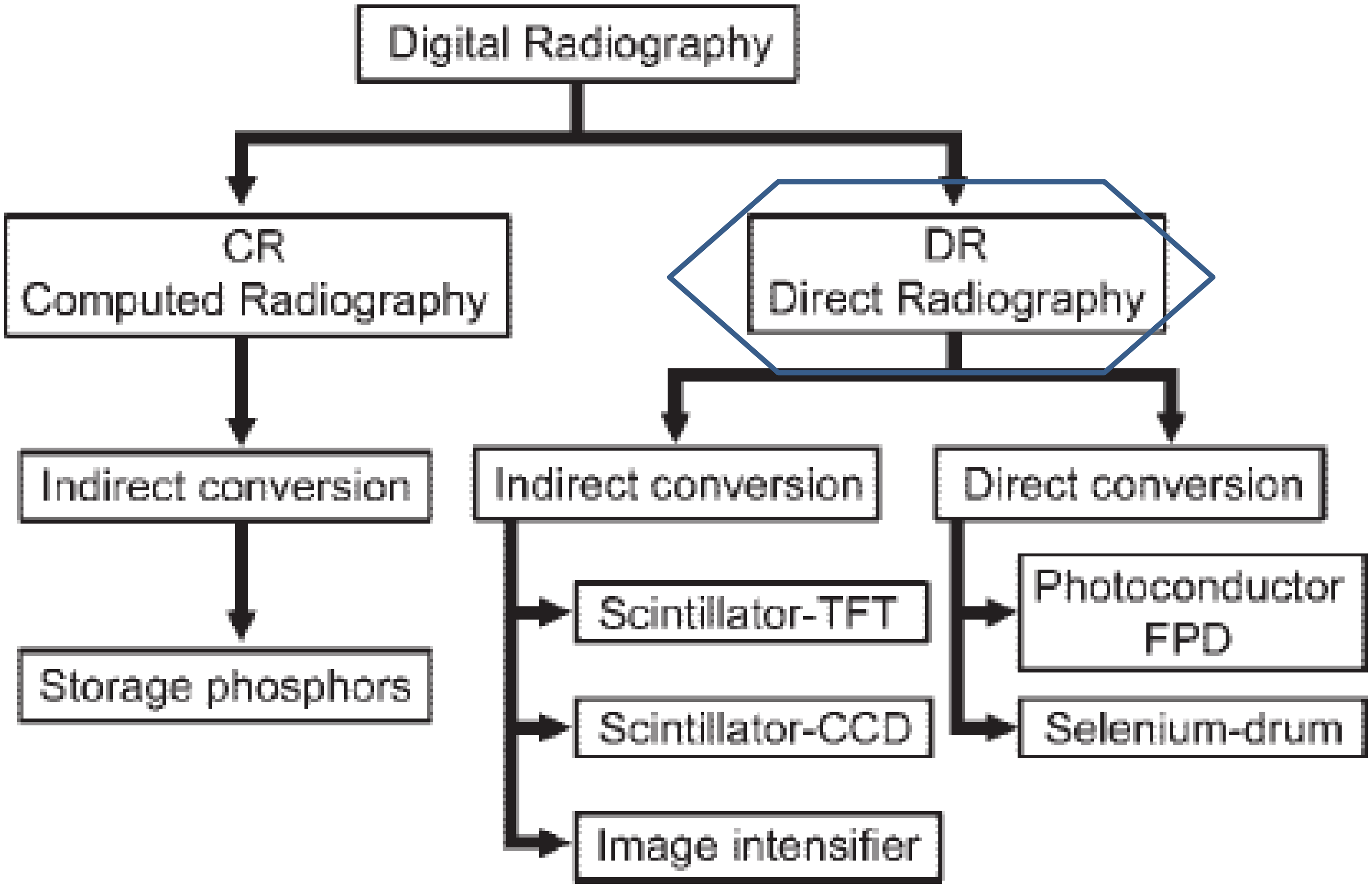
3- If single image plate shows defect → can be easily replaced by the radiographer with no need to change equipment

- Disadvantages:

1- spatial resolution is lower than screen film , yet equivalent diagnostic value

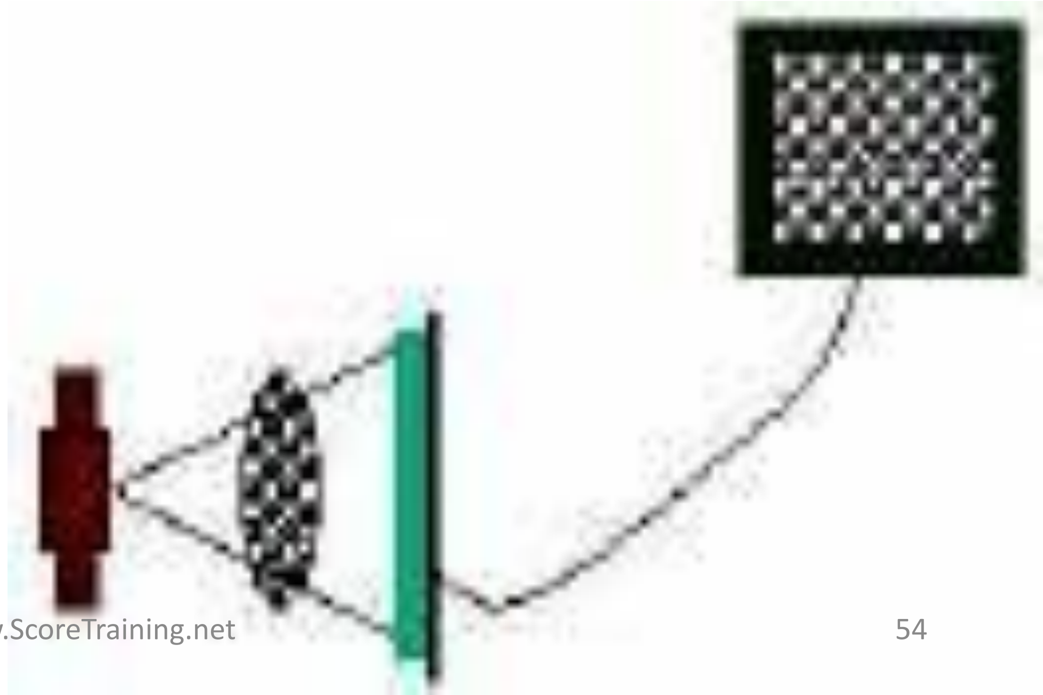
2- image quality and diagnostic value is less than DR

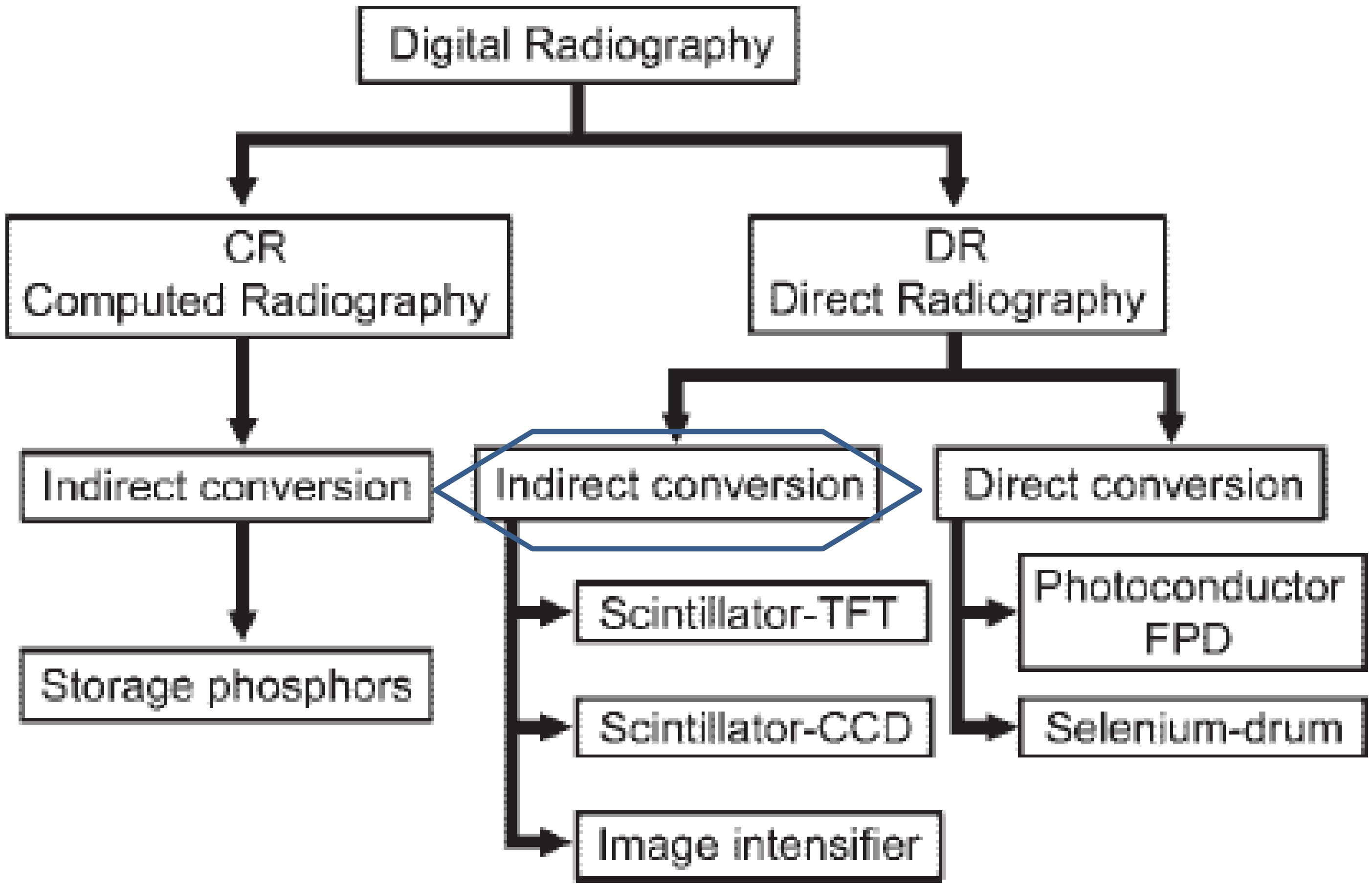
- Used in
  - General diagnostic & mobile radiography
  - Mammography
  - Dental (with small plates!)


Digital Radiography
CR
Computed Radiography
DR
Direct Radiography
Indirect conversion
Storage phosphors
Indirect conversion
Direct conversion
Scintillator-TFT
Scintillator-CCD
Image intensifier
Photoconductor
FPD
Selenium-drum

Direct radiography

- Receptor provides electrical charge output
- No need to reader, No intermediate steps (compare to CR)
- Images available in < 15 seconds
- Much less work for technologist
- Greater throughput

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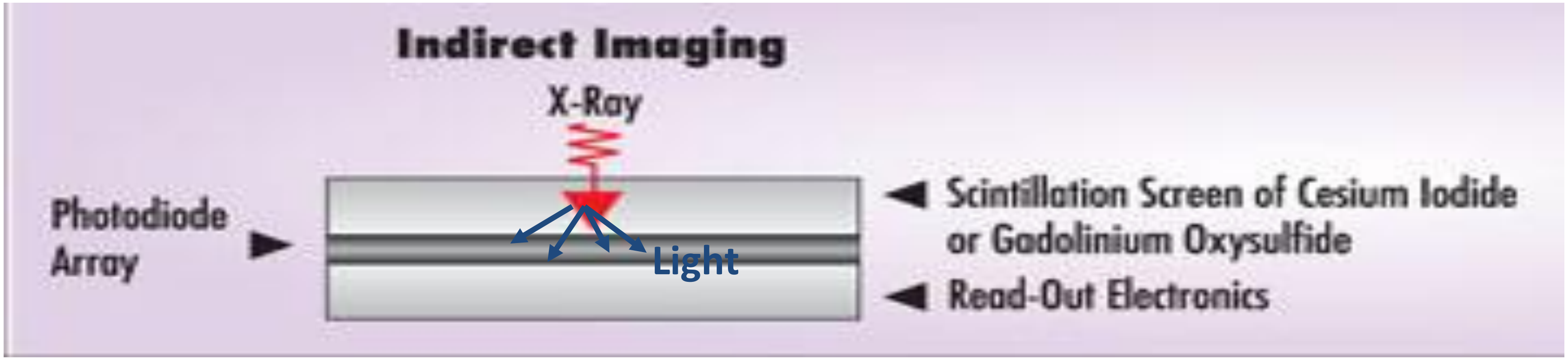
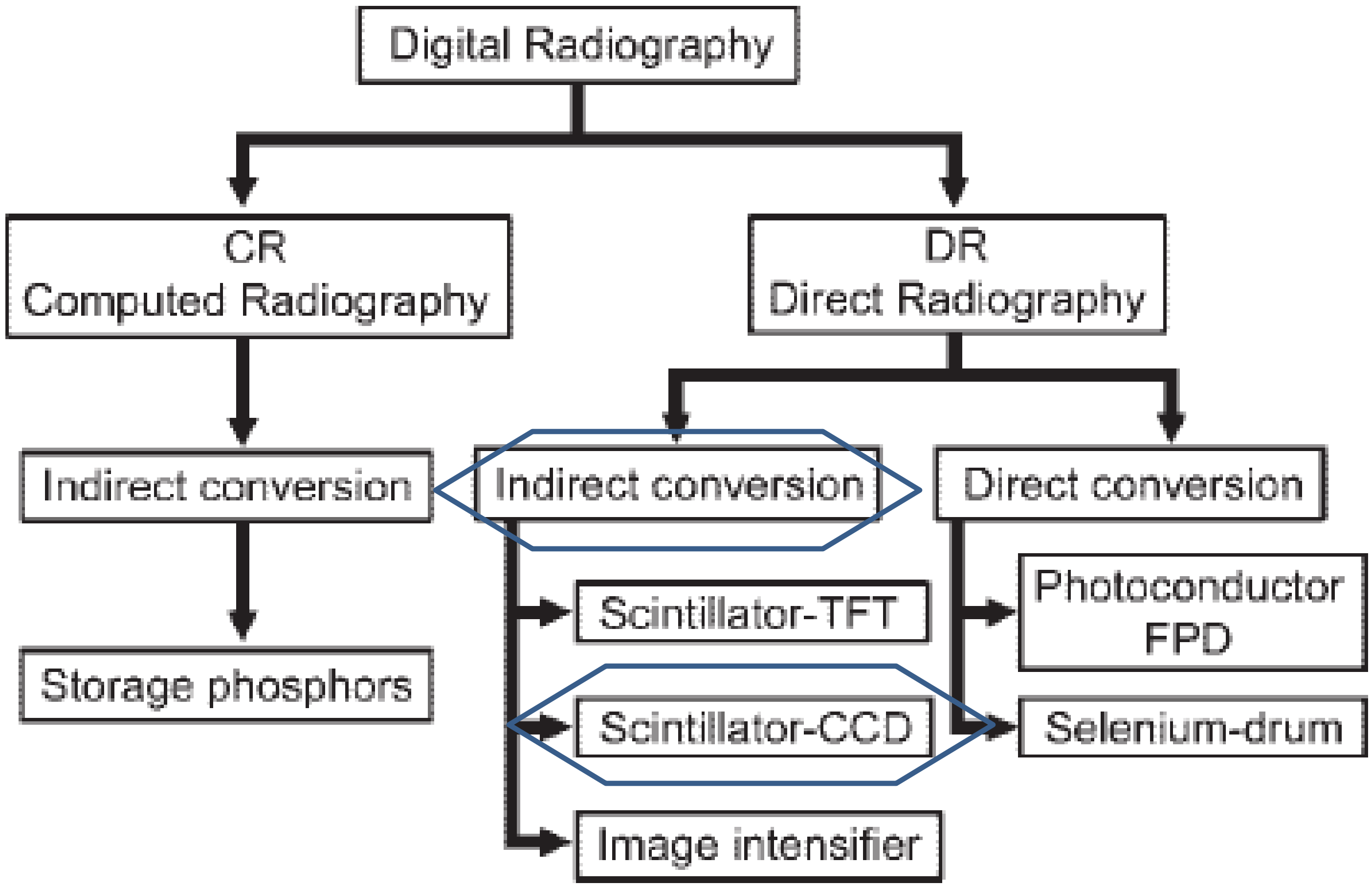


- Indirect DR: x-ray photon → Scintillator → light photons → photodiodes → produce electrons (electrical charge) → readout process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Indirect conversion using CCD (*charge-coupled device*)

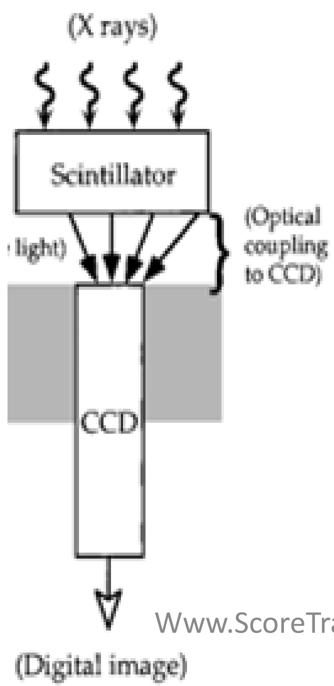
- Scintillator:

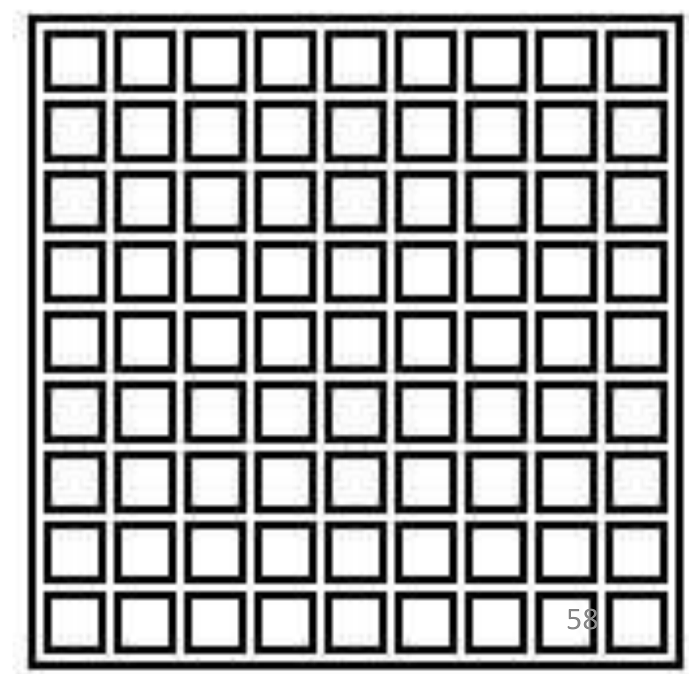
-converts X-ray to light (phosphor)

- Scintillator used is T1-doped cesium iodide

- CCD:

- Made of crystalline silicon (light sensitive)
- Consists of integrated circuit containing array of linked capacitors
- Each capacitor represents a pixel
- Light falls on each pixel , electrons (charge) liberated and build up in each pixel
- i.e. convert light output of the Scintillator to electrical charge



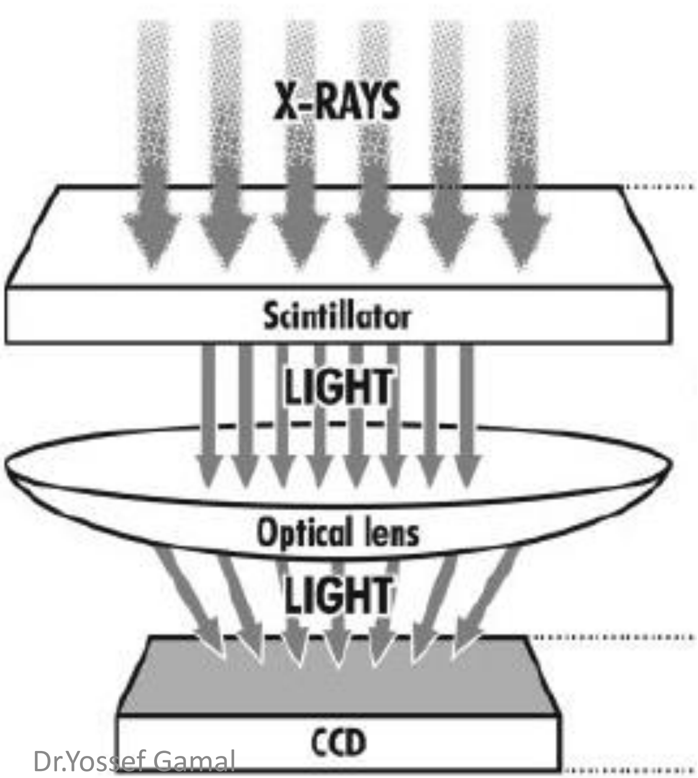


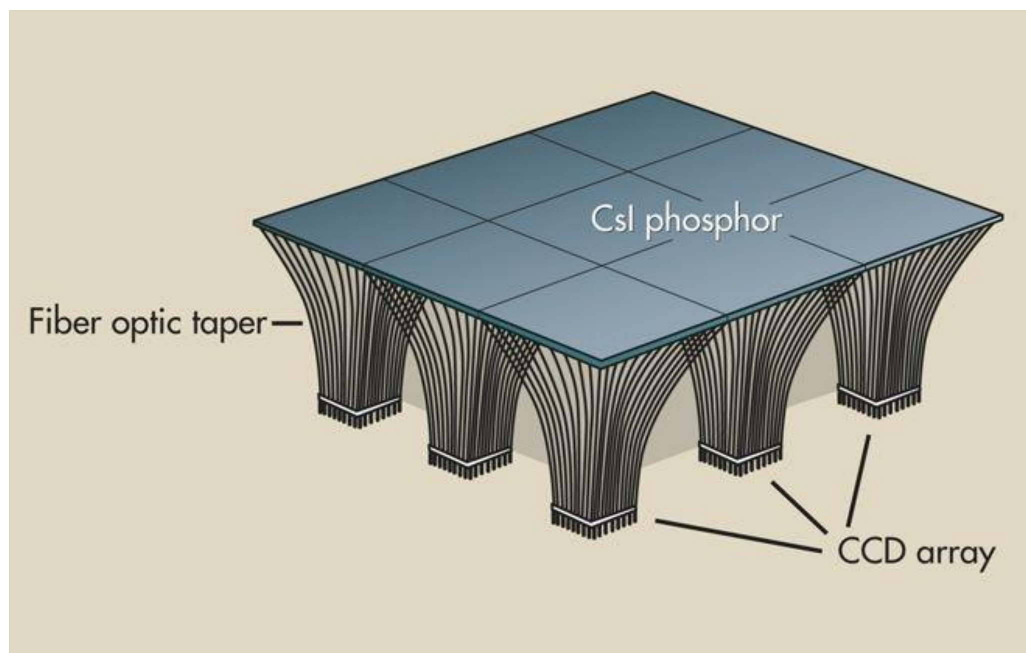
Two types:

<u>1) lens coupled CCD system</u>

- CCD chip can not be manufactured larger than 5 cm^2
- → Array of several CCD chips are combined to create larger detector area
- Optical lens is used to decrease area of projected light to fit CCD array
- Disadvantage:

-the number of photons reaching CCD is decreased →↓ quantum efficiency→↓ S/N ratio,





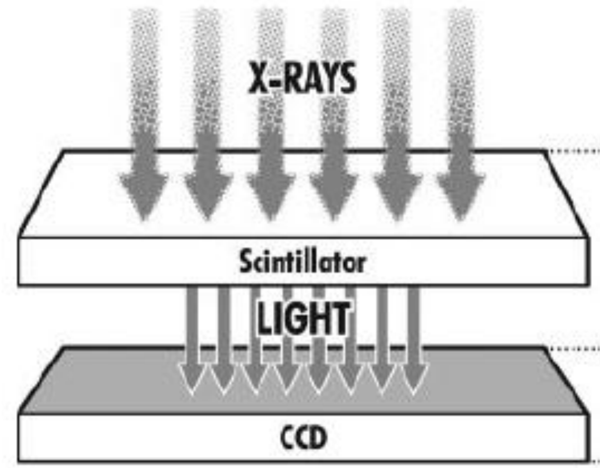
2) Slot scan CCD:

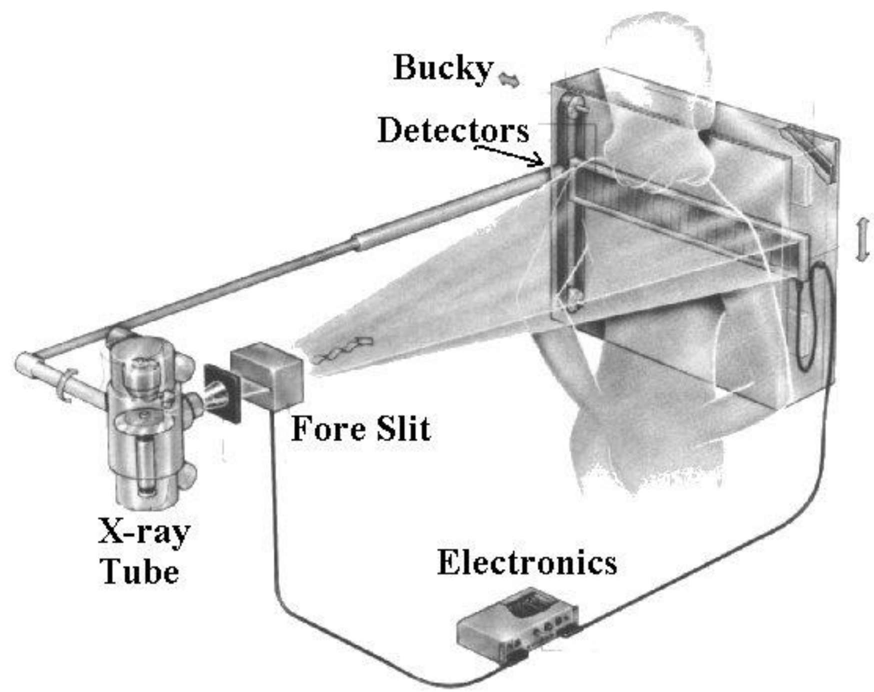
- Collimated fan shaped X-ray beam scan the patient
- CCD detector array of matching width is simultaneously moving
- Advantages:

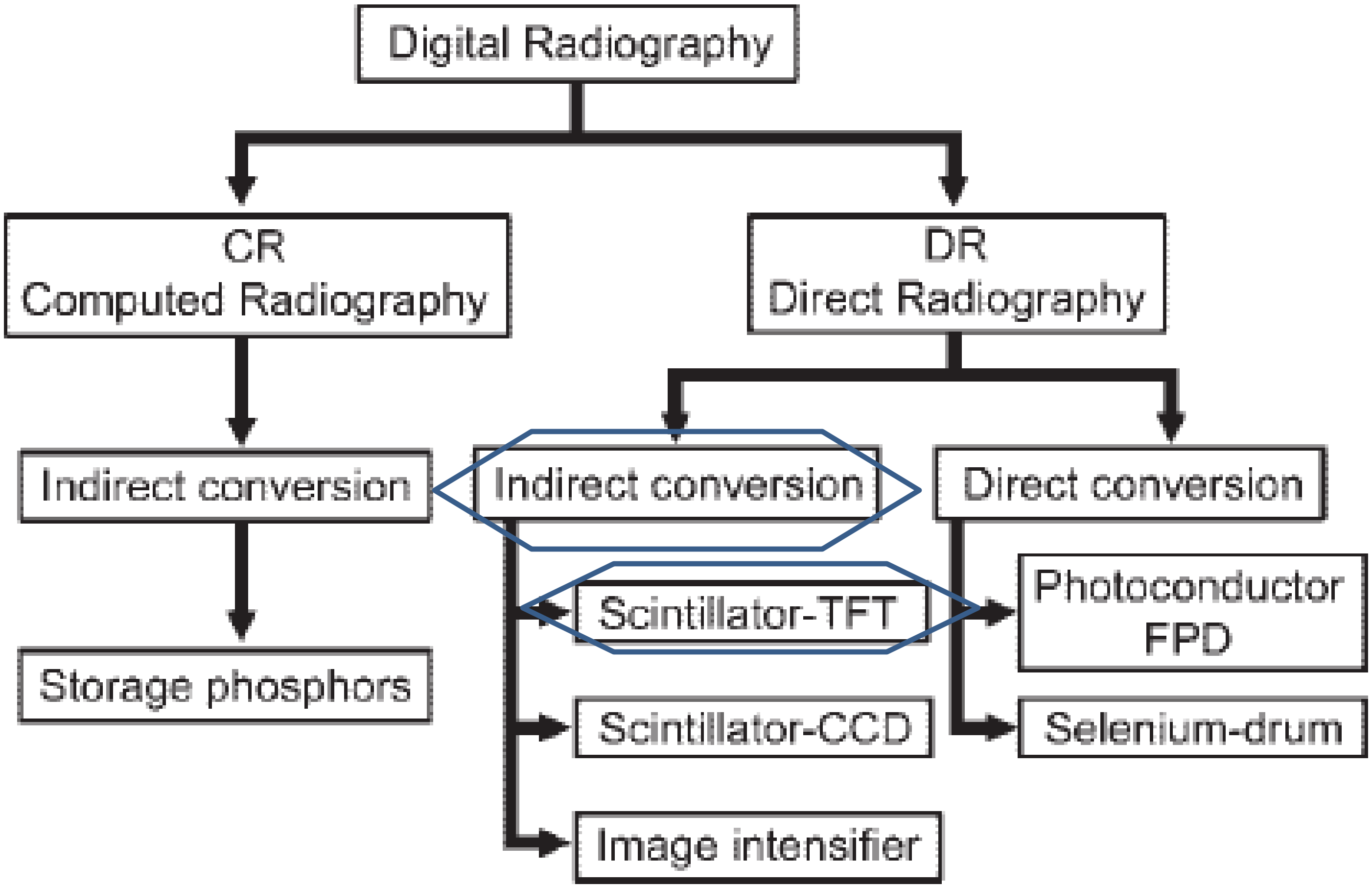
-↓ scatter radiation in the image

-↓noise : compensate the relatively low quantum efficiency of lens coupled CCD system

- Disadvantages: needs fixed installation → used in chest , mammo and dental films
- N.B: Lens coupled CCD is lower than slot scan system in quantum efficiency& S/N ratio



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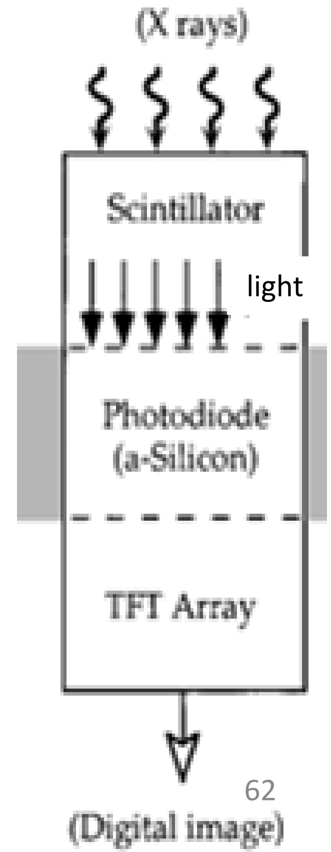
Storage phosphors
Image intensifier

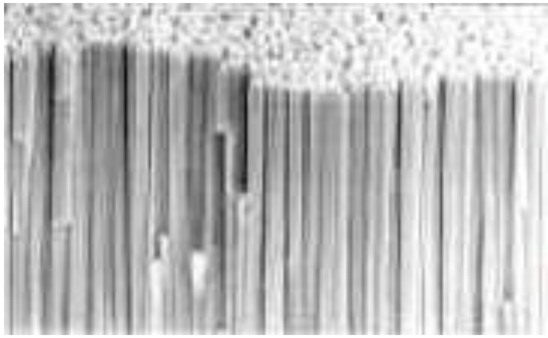
Indirect conversion using flat panel detectors

- Flat panel detector is formed of:

1) Scintillator
2) Amorphous silicon photodiode circuit layer
3) TFT array (*thin-film transistor)*



1) <u>Scintillator layer:</u>



Emit visible light when exposed to X-ray

Types:

A- Cesium iodide (CsI):

Crystals are shaped into 5-10 µm wide needles , arranged perpendicular to the detector surface

This structured array decrease light diffusion in the Scintillator layer → thicker scintillator layer can be used with:

1- increase of absorption efficiency without loss of spatial resolution

2- increase intensity of the emitted light → ↑ quantum efficiency (highest) and better optical properties

Disadvantages: highly vulnerable to mechanical load (fine structure)

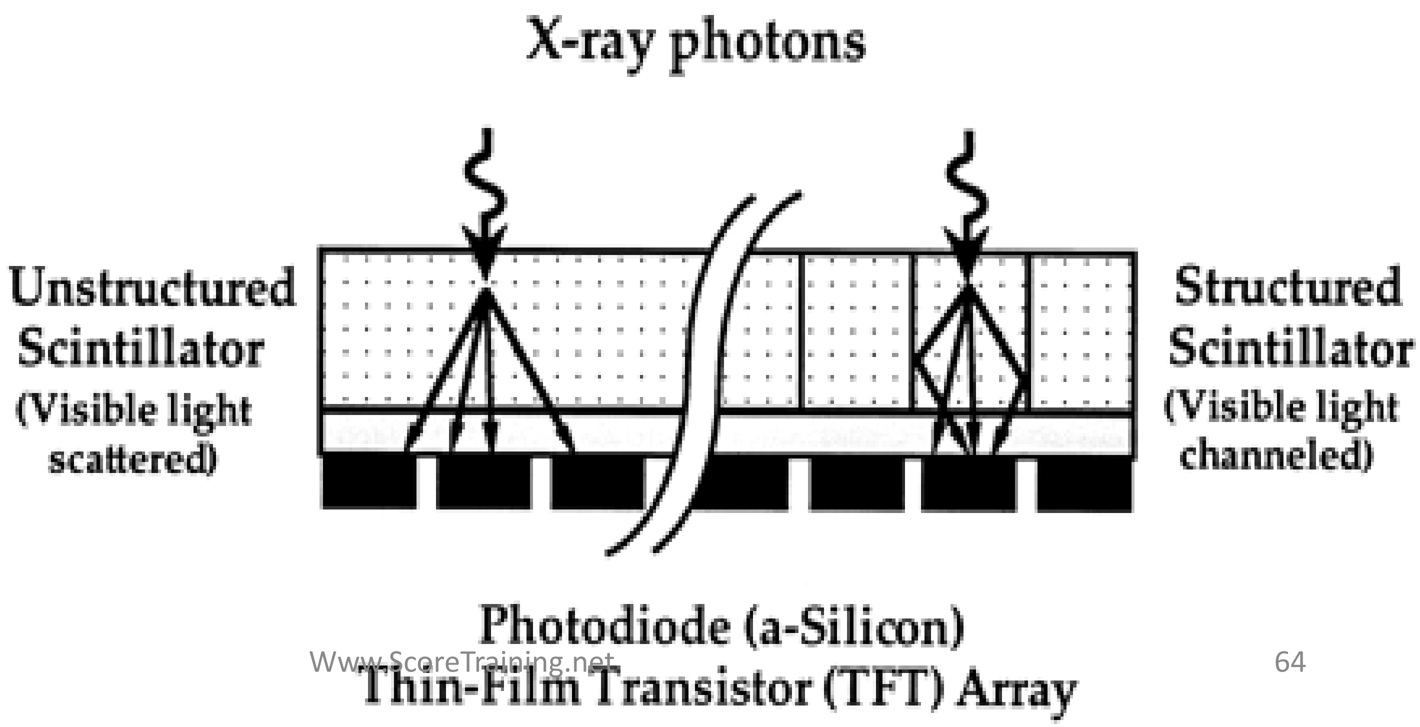
B- gadolinium oxi-sulfide:

- Unstructured Scintillator
- Advantages :

- Cheaper
- Resistant to mechanical stress → used in portable flat panel detector system

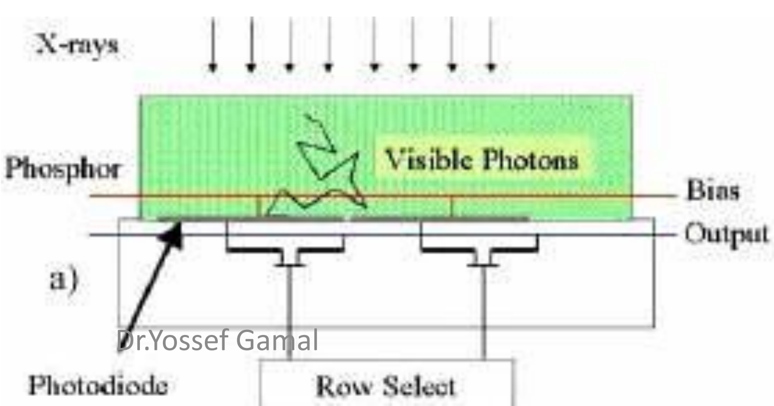
Disadvantages:

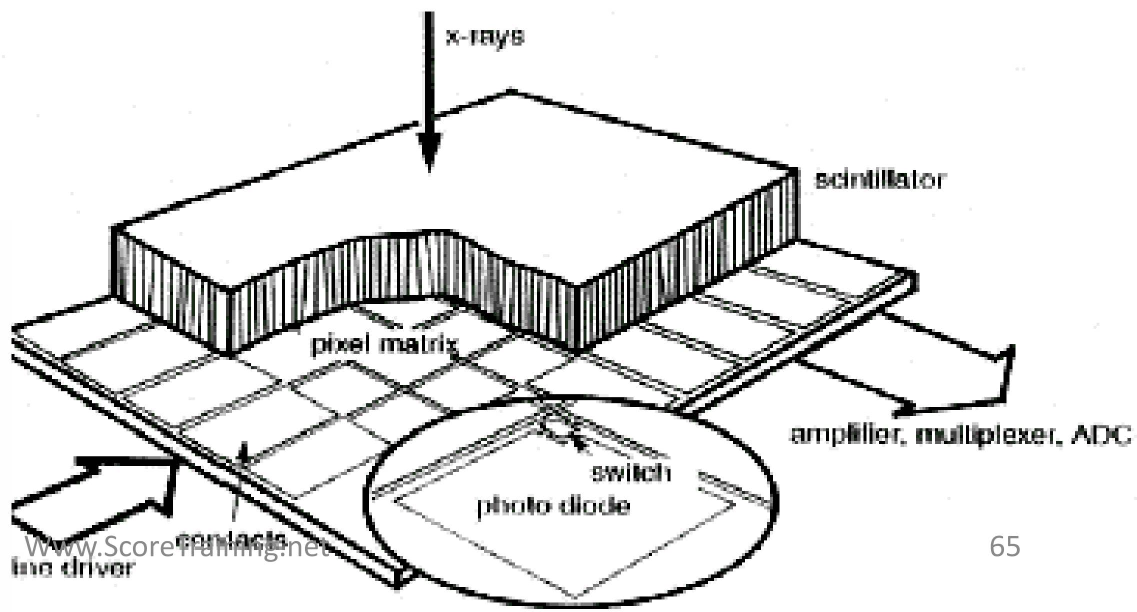
- Lower quantum efficiency



2) Amorphous silicon photodiode circuit layer connected to TFT array:

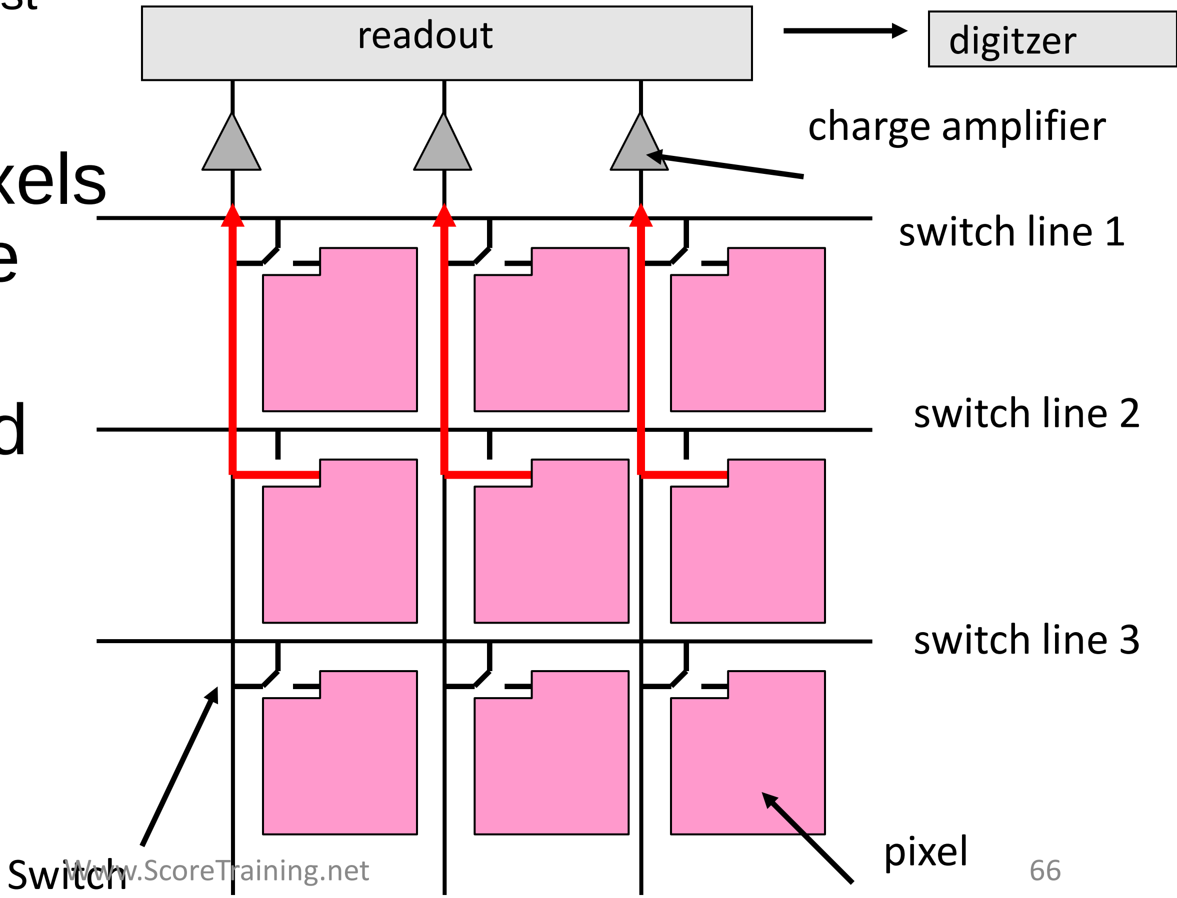
- a made up of small (about 100 to 200μm) detectors (pixels). Each pixel contains a photodiode
- Amorphous silicon photodiode Converts light to electrical charge
- During exposure charge is built in each TFT pixel
- After exposure is complete , charge in each detector element is released by applying high potential (read out)

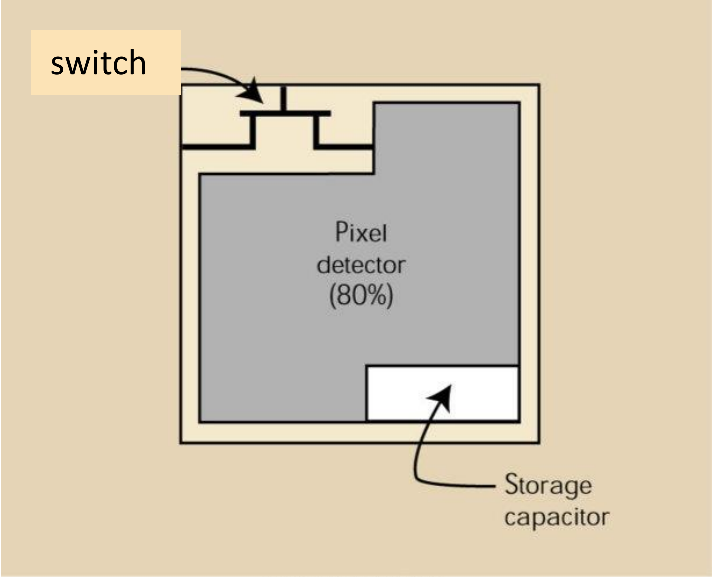




TFT Readout

- All switches are turned off while the x-ray exposure is made
- At the end of exposure, the switches in the 1st row are turned on
- Charges from the pixels are transferred to the charge amplifiers
- Charges are digitized and stored
- row are switched off and the next row is addressed





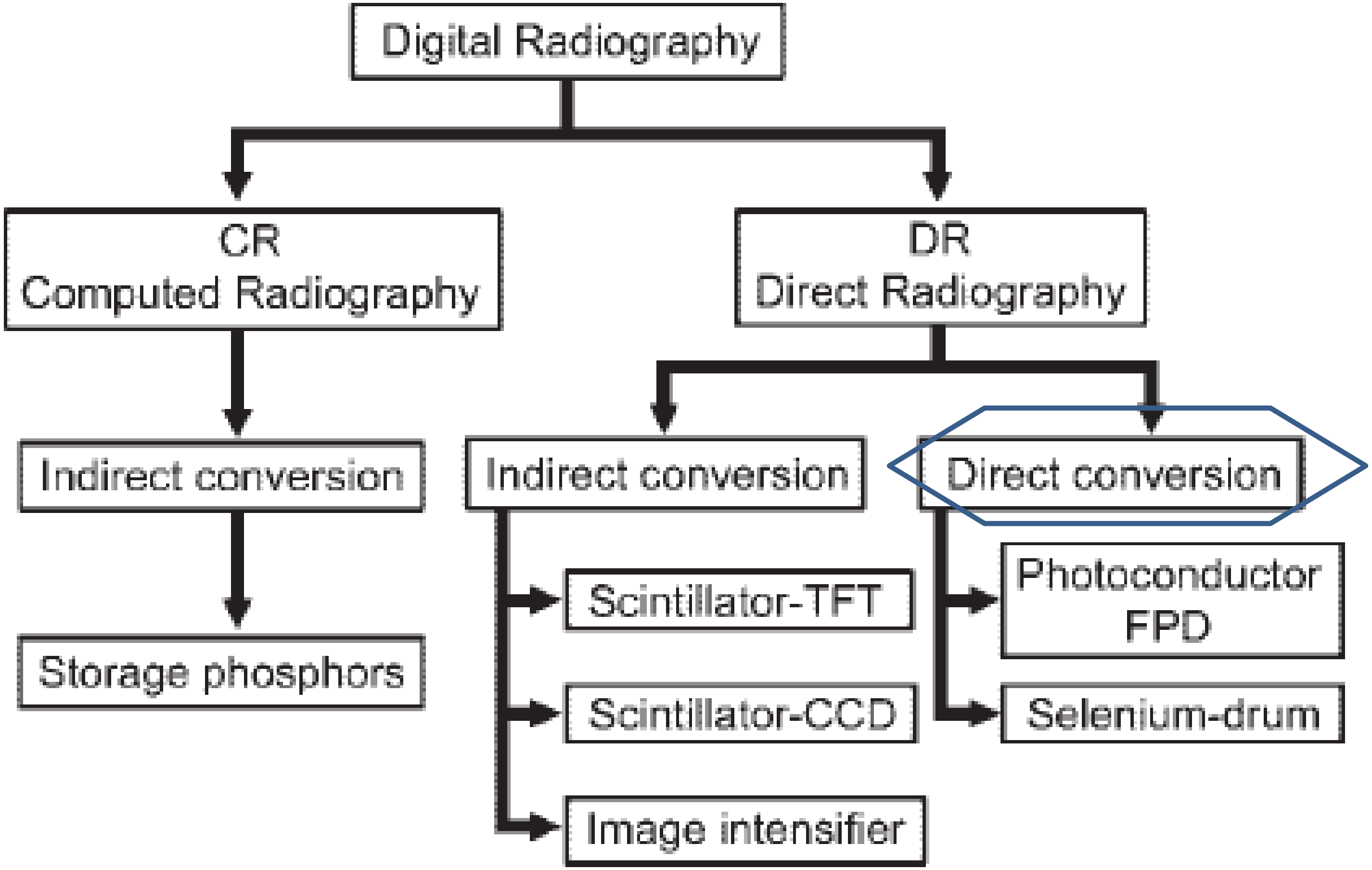
- Fill factor
  - Not all of the pixel is sensitive to signal as a portion of the pixel is taken up by the switch and the storage capacitor
  - The fraction of the pixel that is sensitive to the signal is called the geometrical ***fill factor***

Advantages of Flat panel detector indirect DR:

- Small size : can be integrated in existing bucky tables and thorax stands
- Image generation is almost real time process (time between exposure and image display is less than 10 sec.)
- High throughput
- High image quality

Disadvantages:

- Any detector defect →complete system breakdown (contingency image device is necessary)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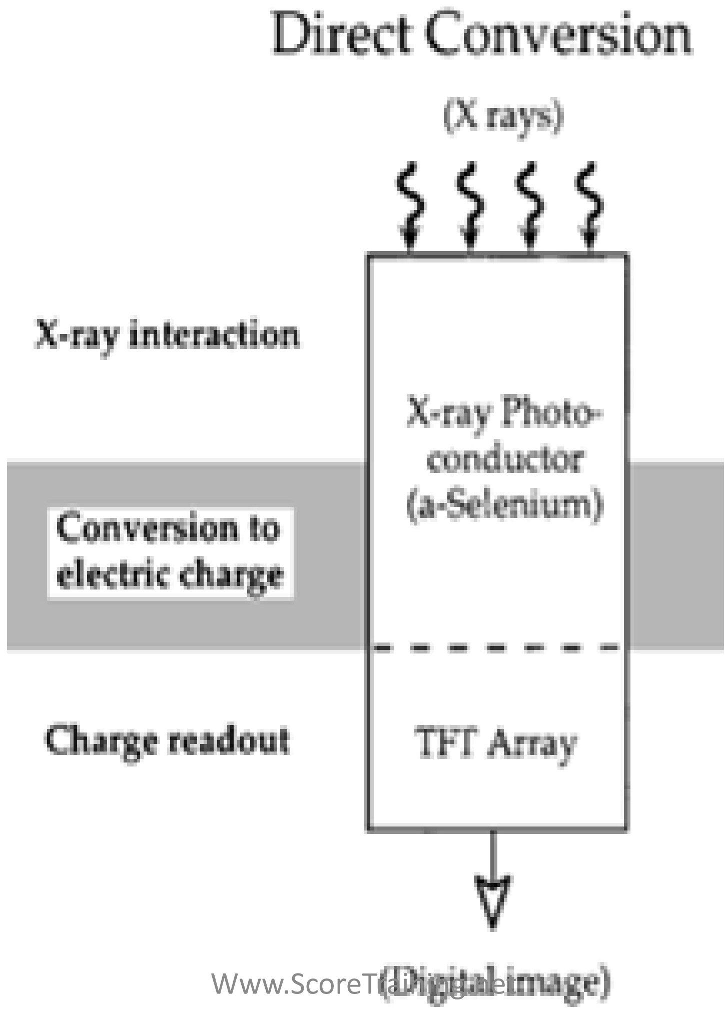
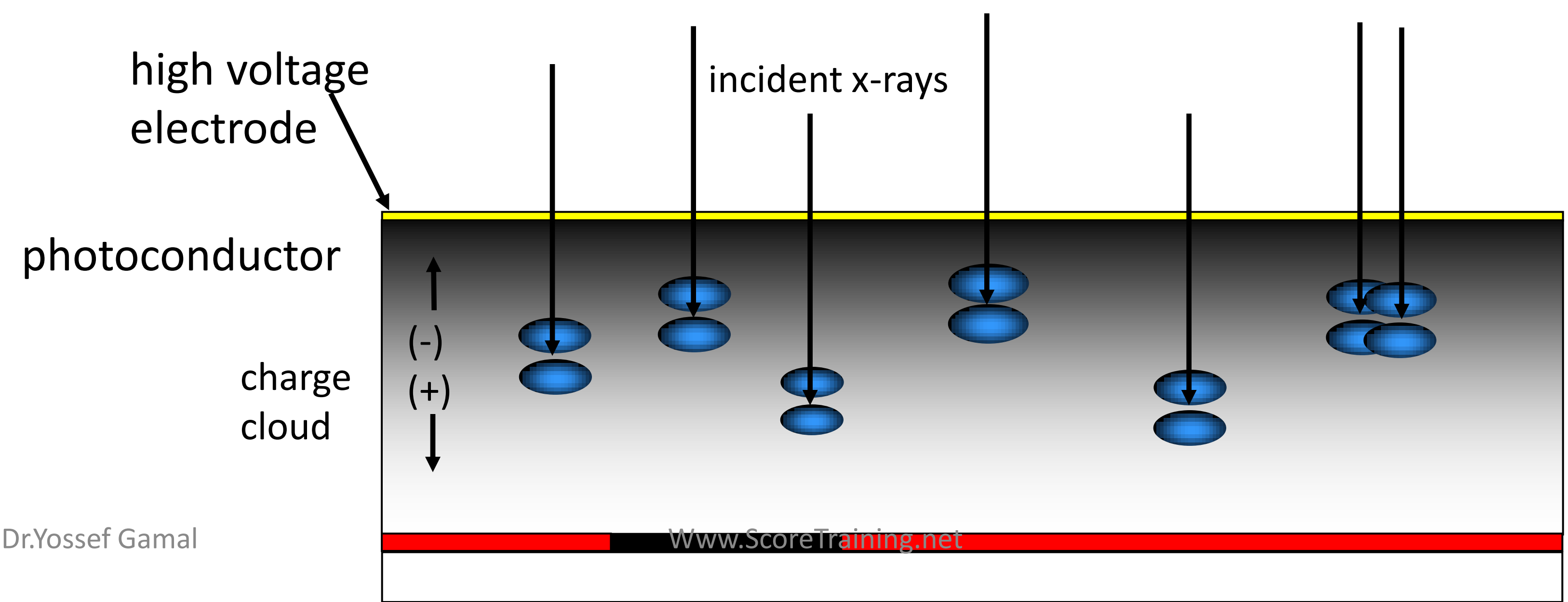
Idea of direct conversion:

Direct-to-Digital Radiography (DDR)



Photoconductor

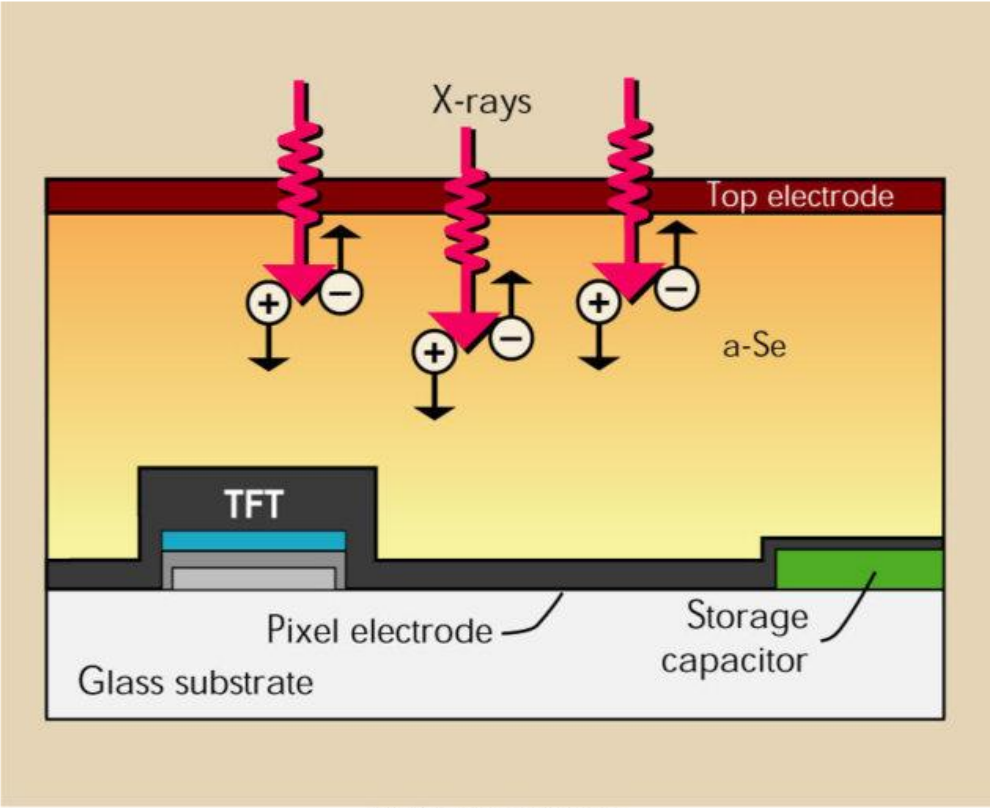
- **Type of detectors which convert X-ray photons into free electrons , which cause electrical charge**
- **Most important : amorphous selenium**
  - Z= 34 : low compared to a typical phosphor → need thick layers for good absorption efficiency
  - Uniform to a very fine scale (amorphous) → The resolution is independent of its thickness
  - Free of the structure noise seen in most phosphors
- **Other types: Lead iodide ,Lead oxide, Thallium bromide, Gadolinium compounds**



- Process:
  - **X-ray generates electrical charge (free carriers)**
  - **electrode is attached to both sides to apply Electric field across the photoconductor to guide the charge**
  - **The greater the field strength, the quicker the charges travel to the electrodes, the less is the time for lateral spread in the material the more is the resolution**
  - The surface of the a-Se have small transverse conductivity to maintain spatial resolution

- The electrode contact must prevent charge from the electrodes to enter a-Se layer (*blocking contact*)

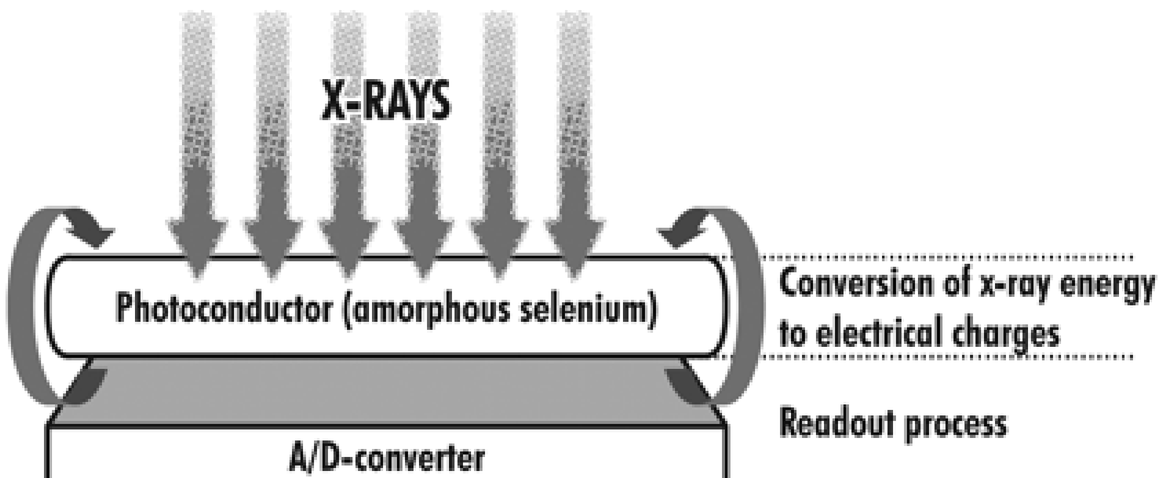
- Pixel size is not dependant on the detector , but on the read out only (highest spatial resolution)



Types of direct conversion DR

1) selenium drum:

- Rotating selenium dotted drum
- +ve electrical surface charge
- X-ray exposure → surface charge pattern α X-ray intensities
- Disadvantage: no mobility due to the mechanical design



2) Selenium based flat panel detectors:

Consists of:

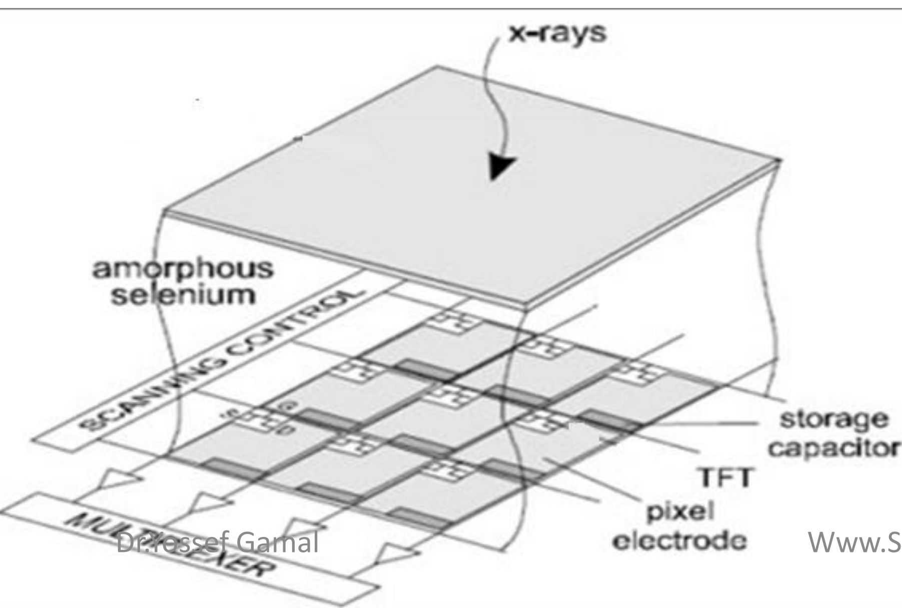
Layer of selenium (convert X-ray to electrical charge)

Underlying array of TFT for readout

Advantages: can be mounted on thorax stands and bucky tables

Used in mammography

N.B: Digital detector function improves as radiation exposure is increased



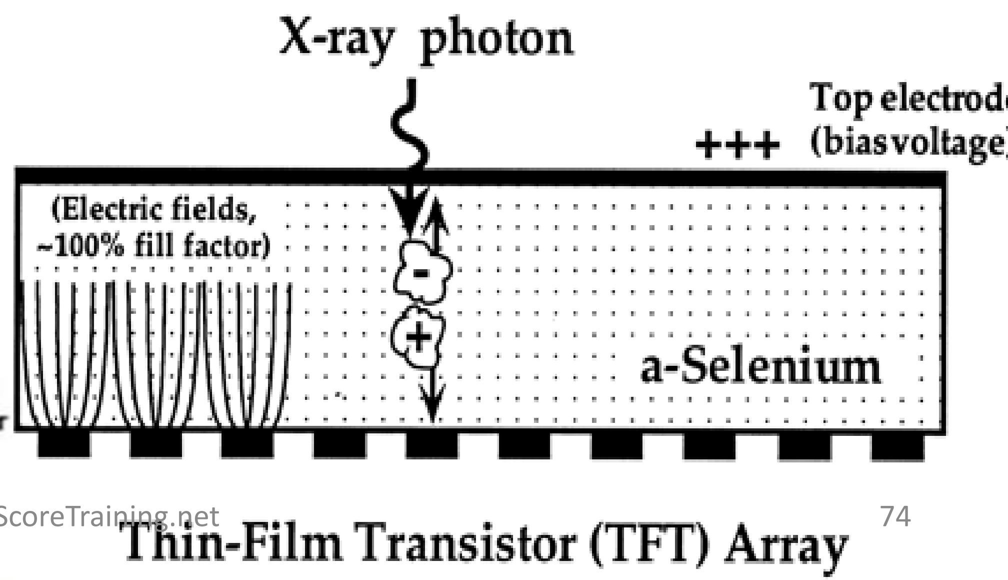


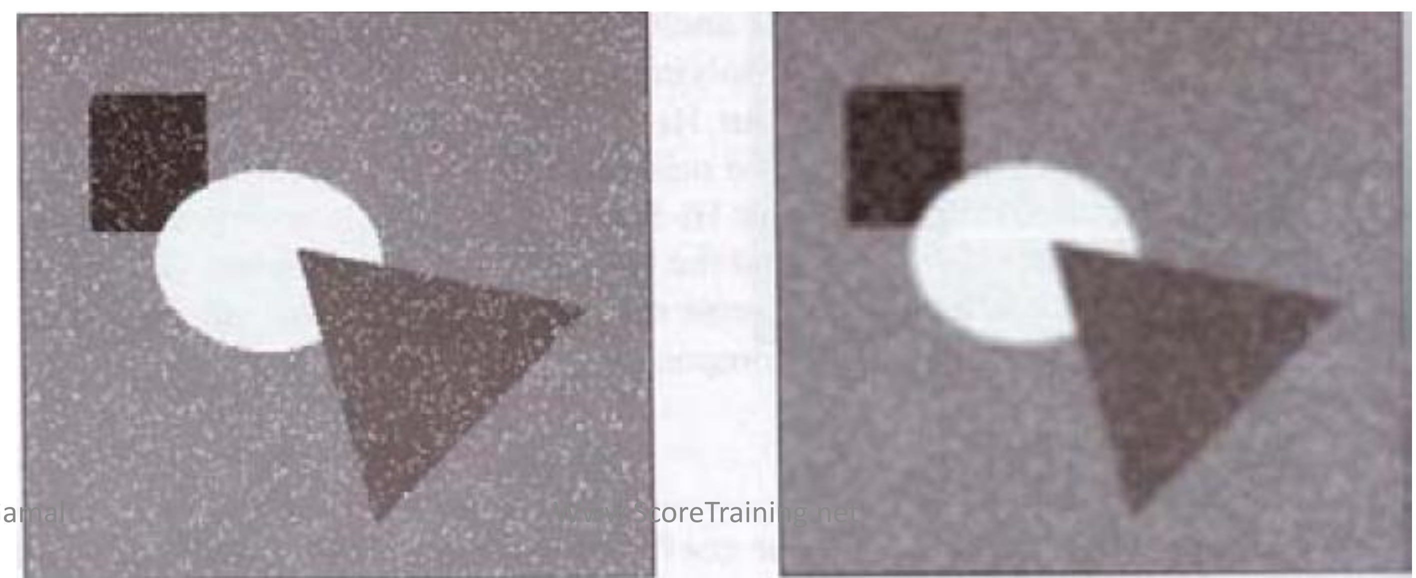
Image post-processing

Image post-processing

- Manipulating readout raw data , examples:

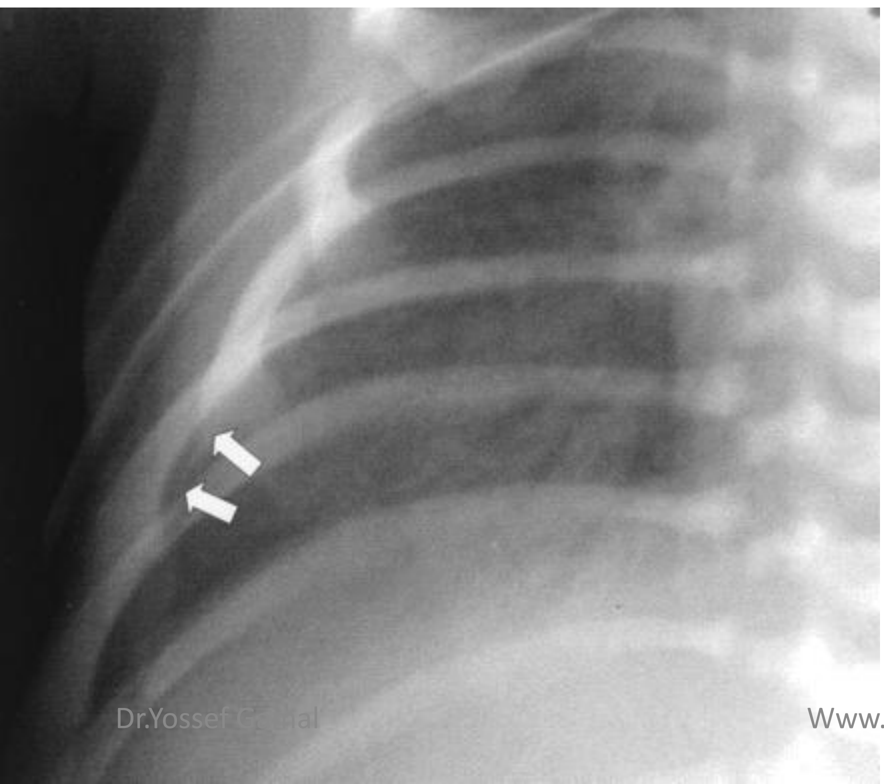
A) Low pass resolution filter:

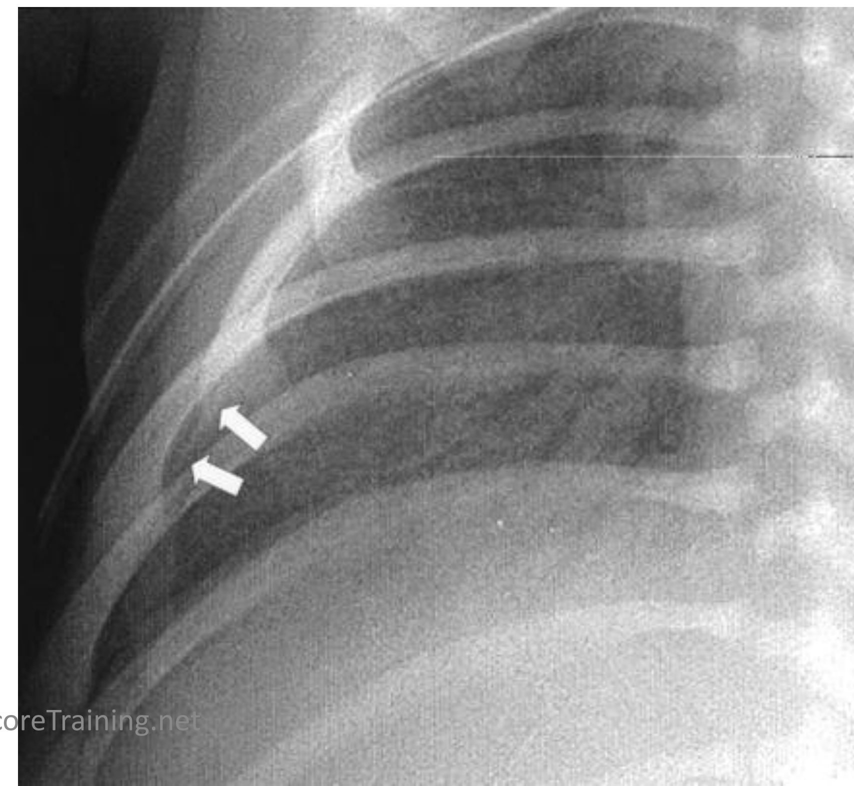
- Each pixel is replaced by the weighted average of the neighborhood pixels
- Effects : smoothen the image → decrease the noise
- Disadvantage: blur some details and edges



B) High pass filter = edge enhancement:

- Increase the difference of grayscale values between neighboring pixels
- Effects: exaggerate the contrast at boundary between structures making the edges more visible
- Disadvantages: increase the noise , may generate false structures in the image





- Notes:

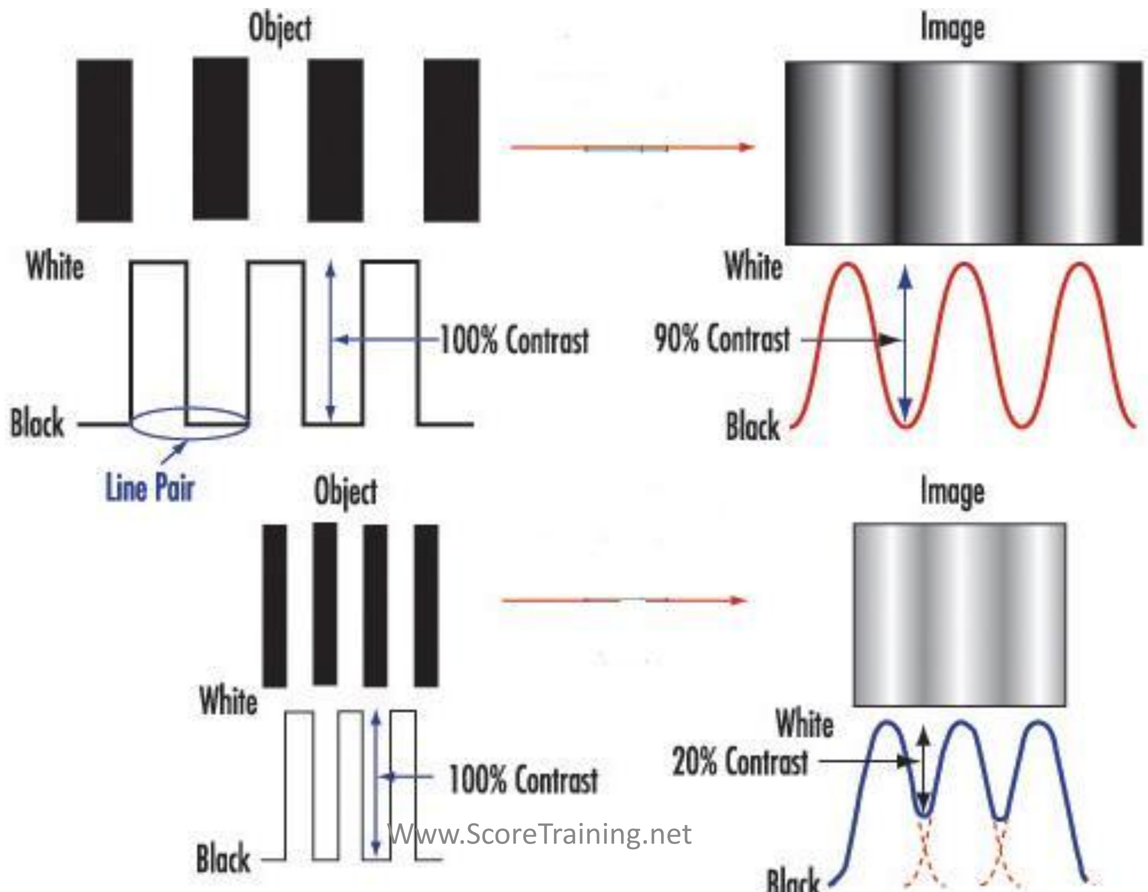
1) Spatial resolution can not be influenced by processing , yet , by changing processing variables (e.g. edge enhancement) low spatial resolution can be partially counteracted
2) Processing algorithms are adapted for each anatomical regions

Modulation transfer function (MTF)

- Definition: ratio of output and input modulation

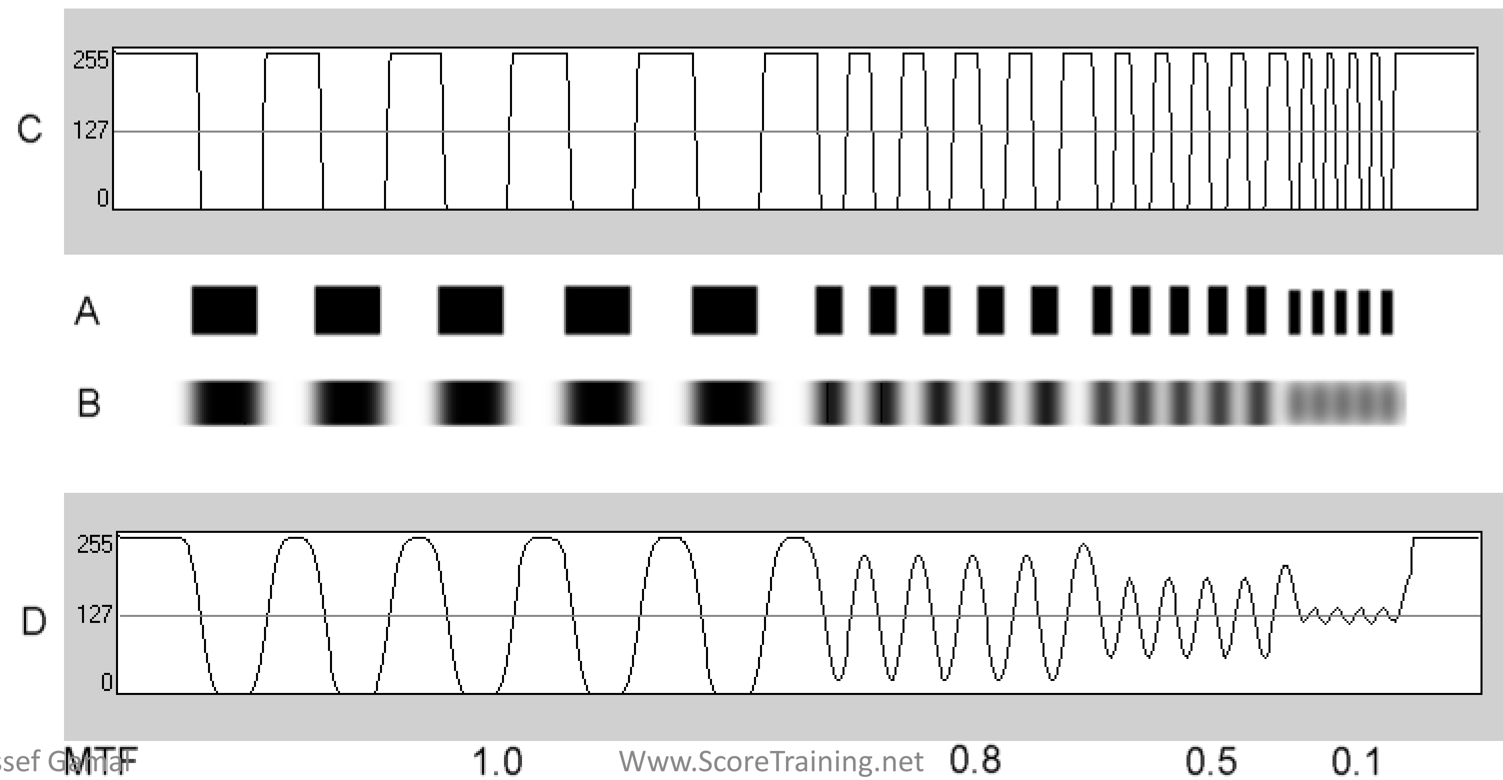
i.e. capacity of the system to transfer modulations of the input signal at a given spatial frequency to its output

i.e. A measure of information loss in the image compared to the object

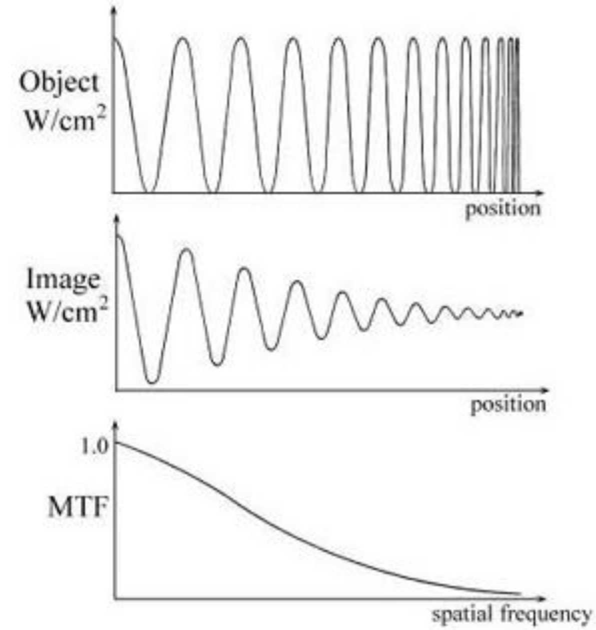


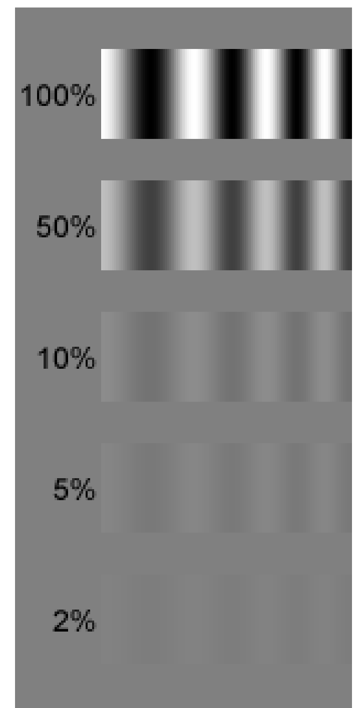
i.e. MTF = amount of blur at certain spatial frequency (↑spatial frequency →↑blur)

- MTF varies with spatial frequency from 100% at low spatial frequency , towards zero at higher spatial frequencies
- Practical limits of MTF : 0-1 (when will it exceeds 1?)

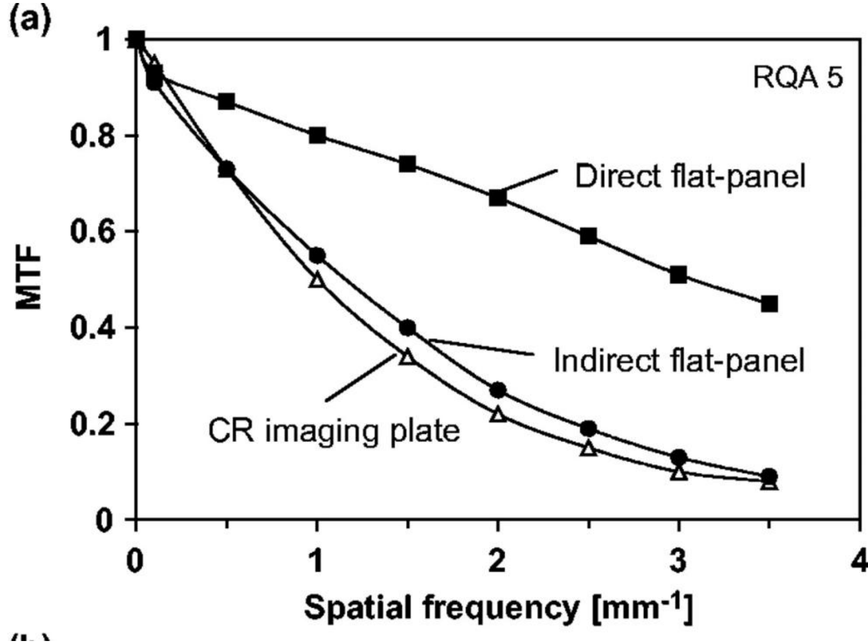


- MTF depends on the system and the spatial frequency of the area
- MTF of the system = product of its individual components
- Fine structures with sharp edges corresponds to high spatial frequency → require system with high MTF
- A common way to compare two systems is quoting a figure at which MTF = 10% for each
- In general radiology : relevant image details are usually less than 2 cycles/mm
- In mammo: details are › 5 cycles/mm





MTF of Direct flat panel is the best as the electric field in the selenium layer, necessary to separate the charges, inhibits the lateral diffusion of the charge cloud, thus preserving a high resolution or MTF

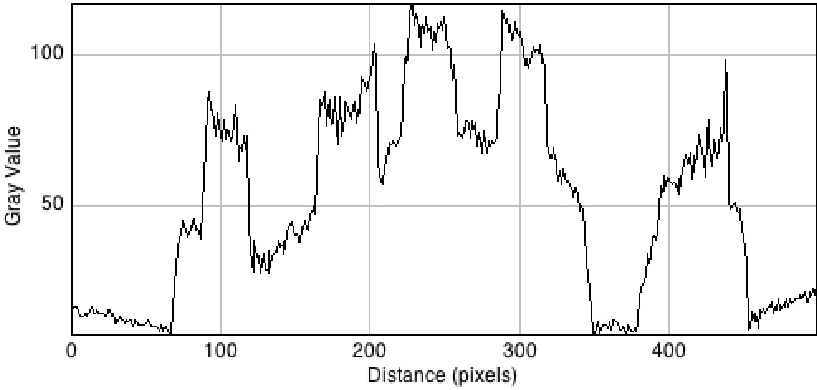


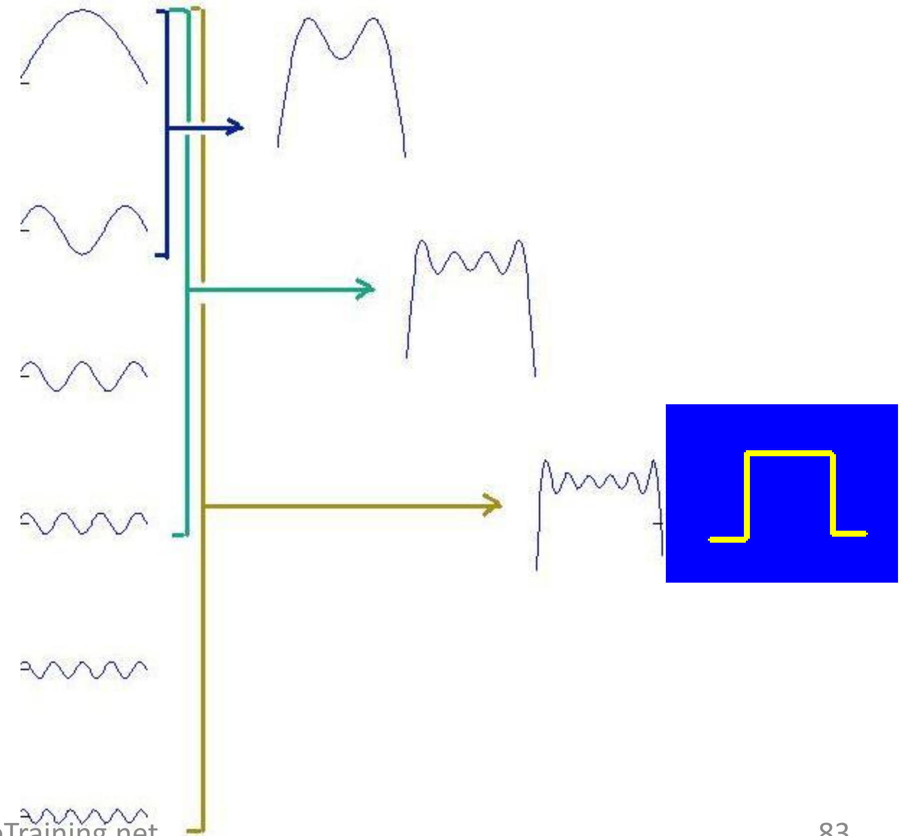
Fourier transform

- Technique of converting from spatial to frequency domain

i.e. image signals are converted to sine waves in terms of spatial frequency and amplitude

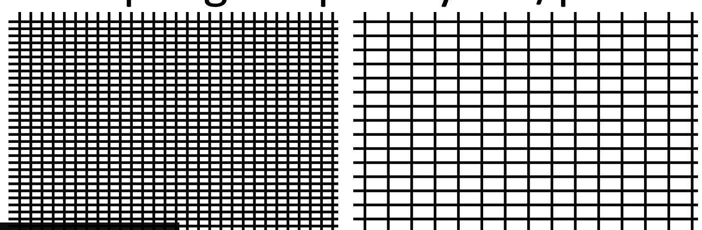
- Fourier methods are used in digital image to enhance the display of image detail

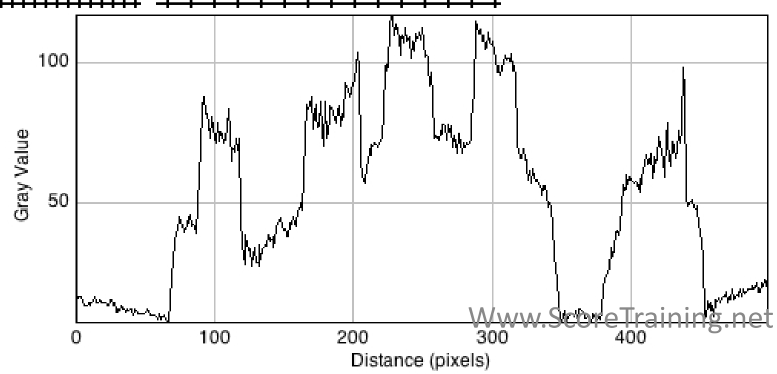


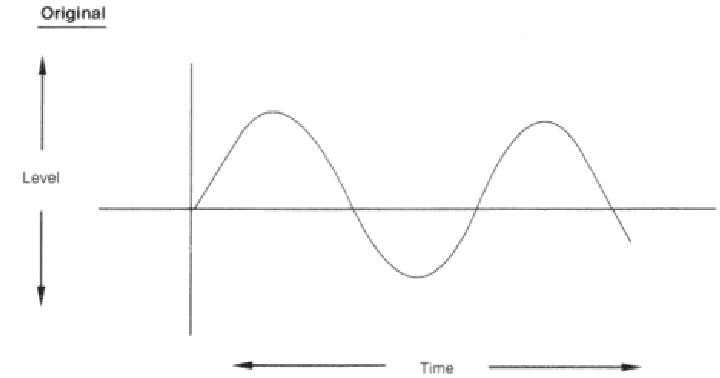


Sampling

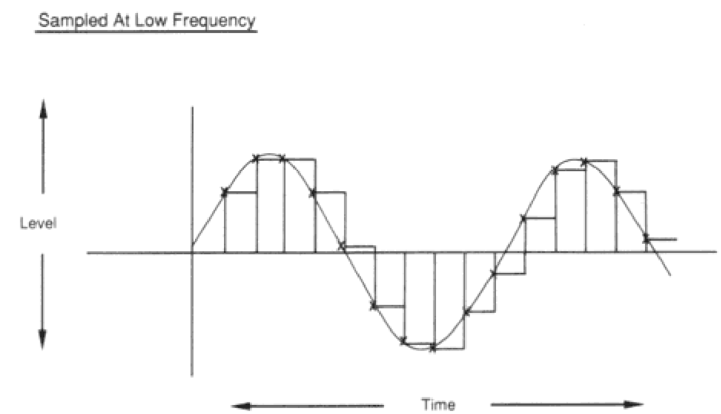
- Image sampling is the process used to digitalize the analogue spatial information in digital image
  - i.e. Process of creation of digital image from the Fourier sine waves of many frequencies
- Sampling frequency:
  - Rate at which the analogue wave is sampled to create digital image
  - Determine the pixel size in the digital images
  - sampling frequency = 1/pixel size



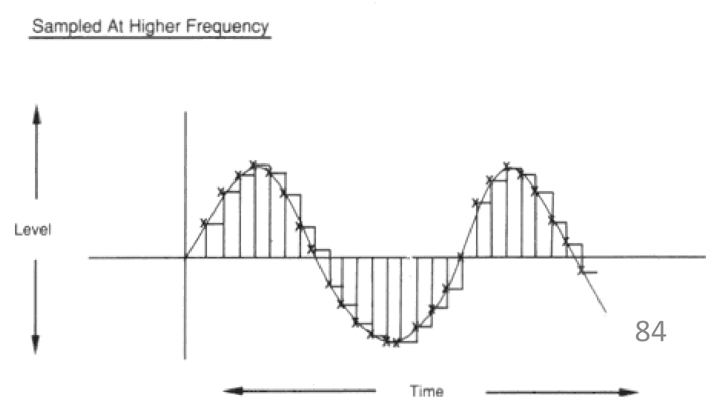




The original wave shows a continuous variation in level with respect to time

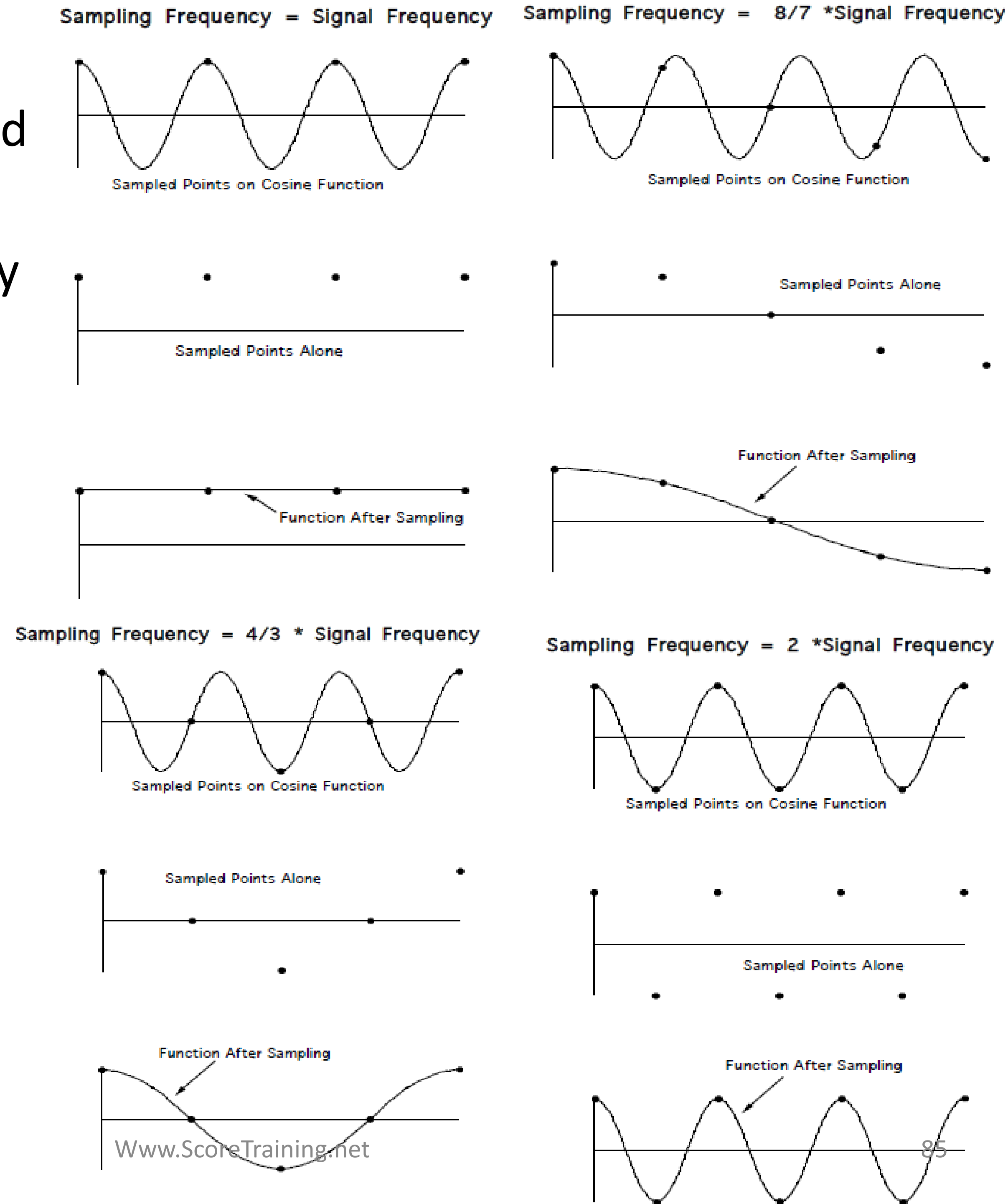


Widely spaced sample points lead to a stepped wave when played back via the DAC

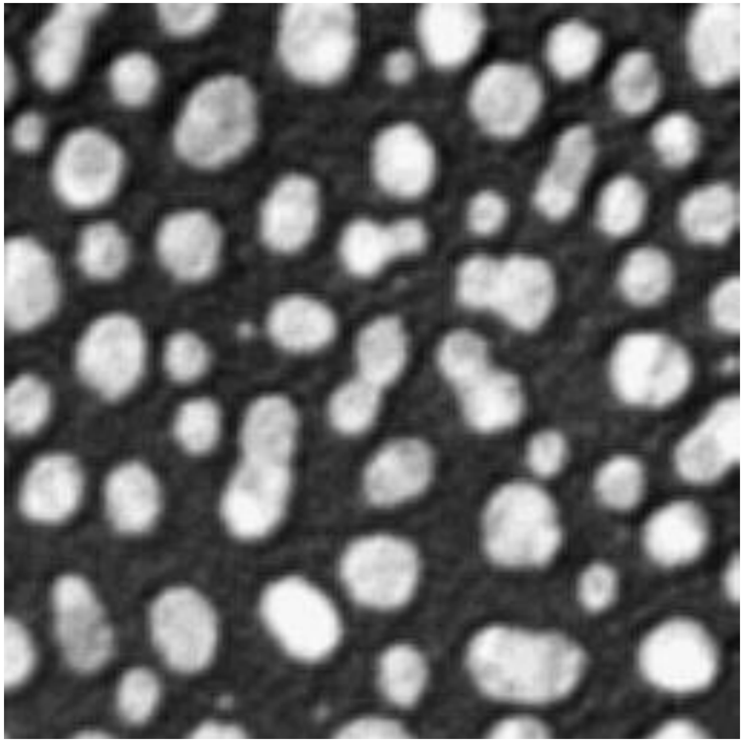


Nyquist law

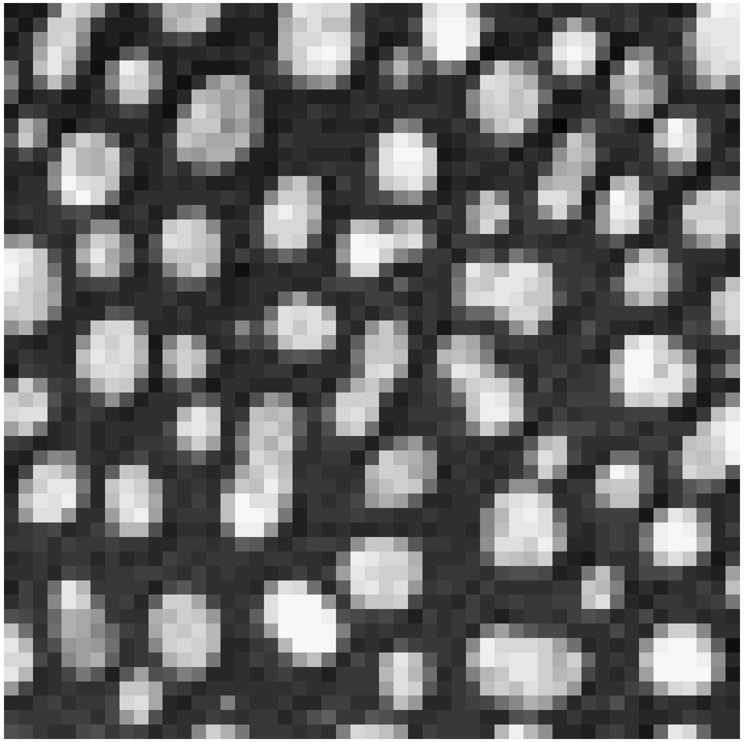
- Signal must be sampled at least twice/cycle
- i.e. sampling frequency must be at least twice of the highest frequency present in the image



Good sampling



under sampling



Nyquist frequency

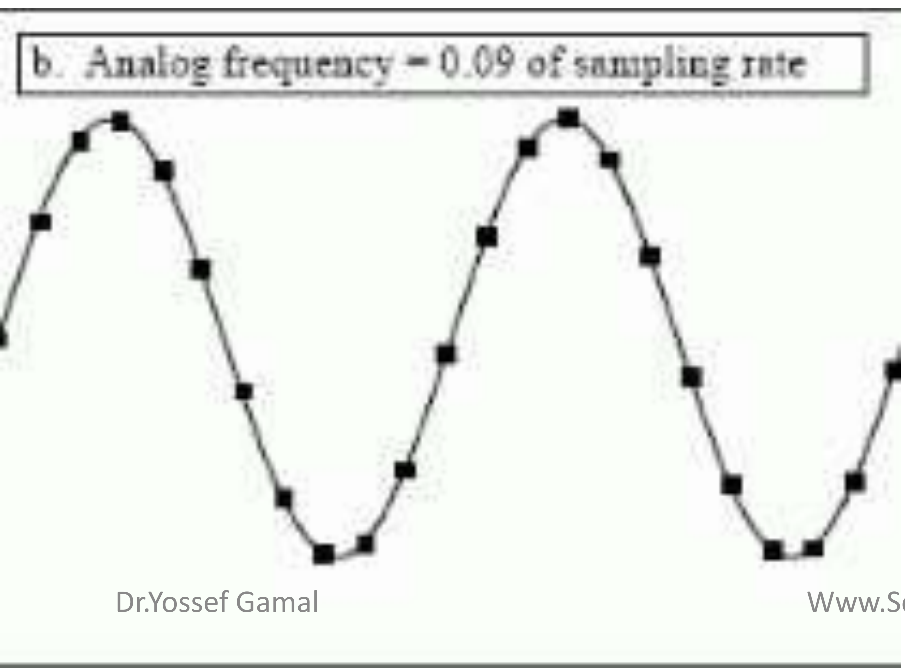
- Maximum signal frequency that can be accurately sampled

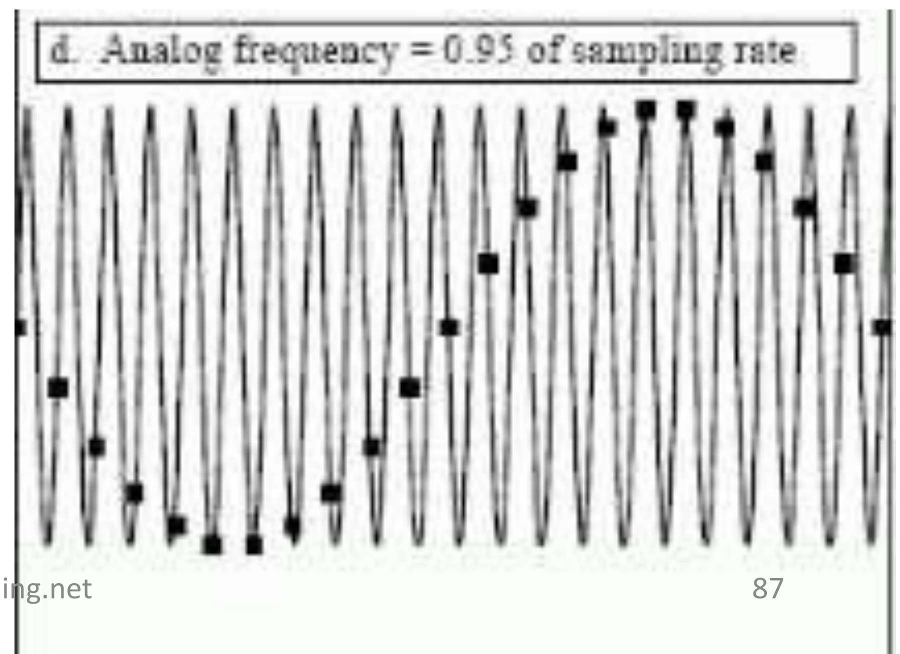
= maximum achievable spatial resolution (in cycles/mm)

$$= \frac{\text{sampling frequency}}{2} = \frac{1}{2 \times \text{pixel size}}$$

e.g. pixel size = 200 µm → maximum detectable spatial frequency = 2.5 cycles/mm

- Aliasing: high frequency signal is faultily recorded low because sampling frequency is ‹ Nyquist frequency





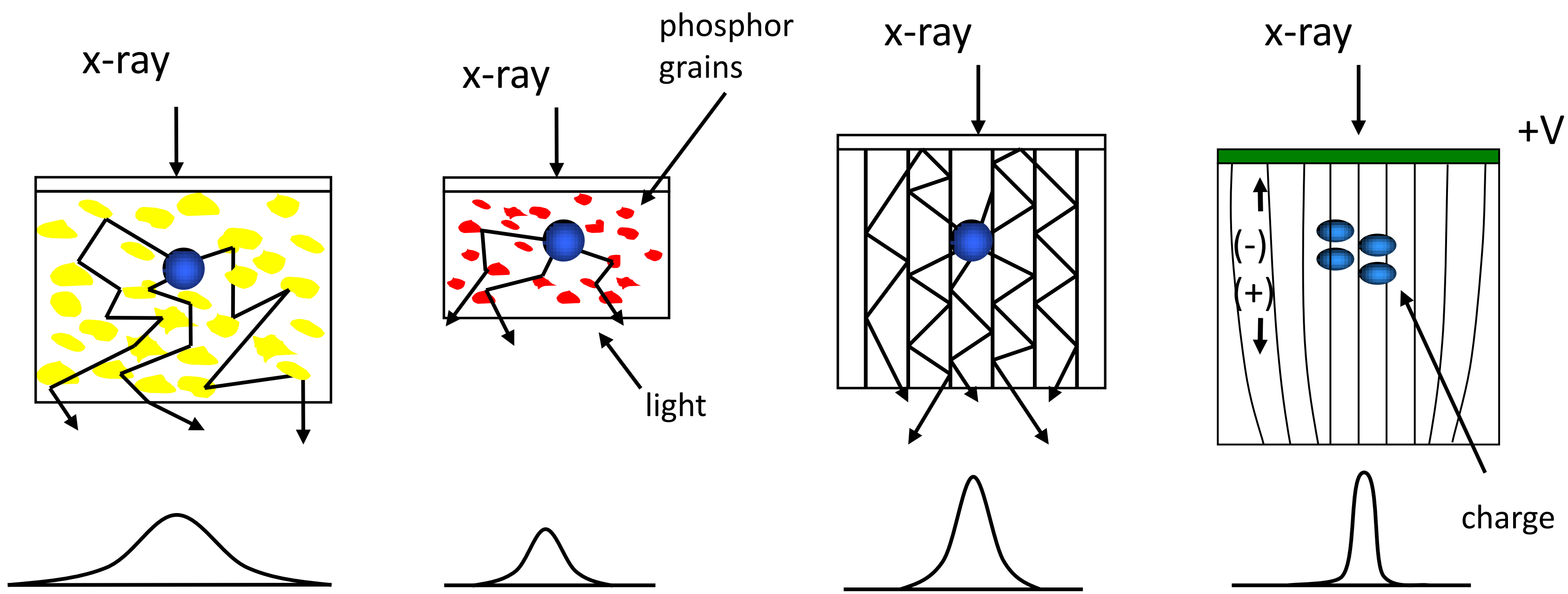
Spatial resolution in digital radiology

- In digital system : it is limited by the pixel size
- Scattering of light photons decrease spatial resolution so that:

1- spatial resolution of direct conversion DR › indirect conversion DR

2- spatial resolution of structured Scintillator › unstructured Scintillator

Spatial resolution of different digital systems



CR | INDIRECT DR | INDIRECT DR | photoconductor

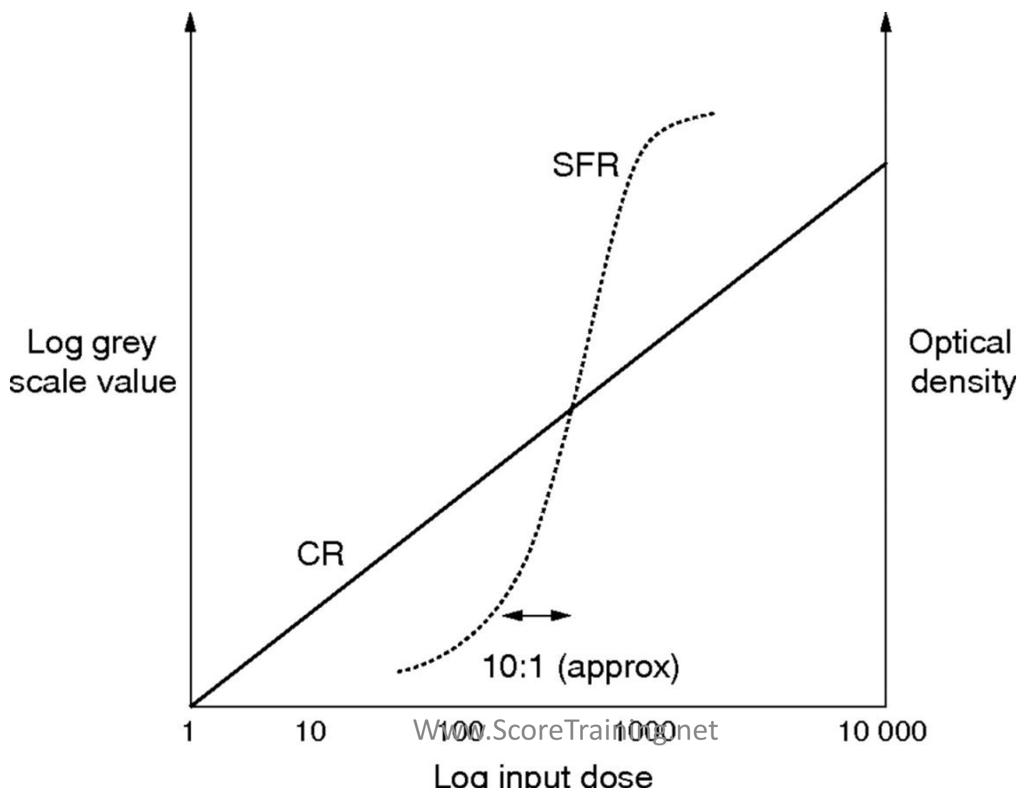
$BaFBr_{0.85}I_{0.15}:Eu^{2+}$ | GdO_2S_2 | CsI:Tl | *a*-Se

Detector Dynamic range

- Range of X-ray over which meaningful X-ray image is obtained = latitude
- Digital detectors have very wide dynamic range
- Results:

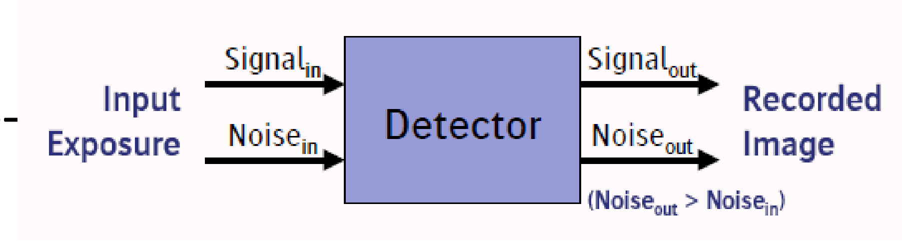
1- low risk of failed exposure

2- difference between images (e.g. bone, soft tissue and air)can be seen in one image without need of additional images



Detective quantum efficiency (DQE)

$$DQE = \frac{(\text{S/N ratio at detector output})^2}{(\text{S/N ratio at detector input})^2}$$



- Describes how effectively an x-ray imaging system can produce an image with a high signal-to-noise ratio relative to an ideal detector
- Detector with high DQE =

1- has high photon detection Efficiency:

↑DQE → less radiation needed to achieve the same image quality → ↑ image quality with the same exposure

2- low noise is added to the detected signal

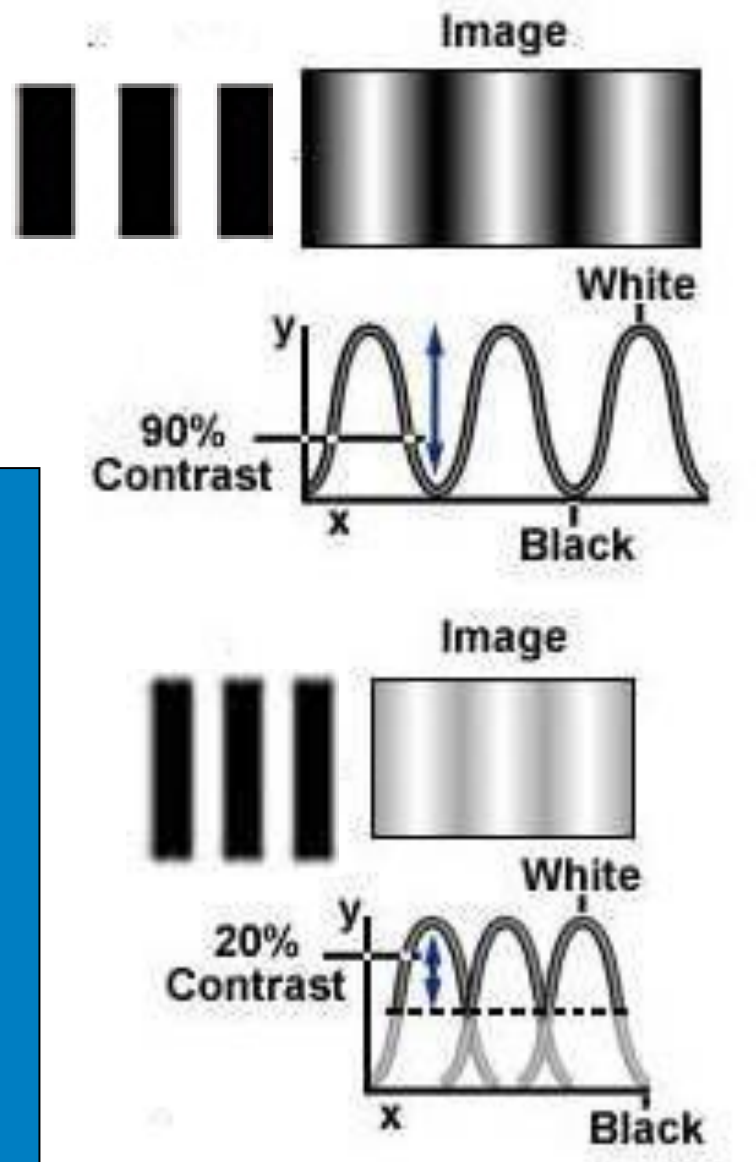
i.e. added to the quantum noise that is present in the X-ray beam

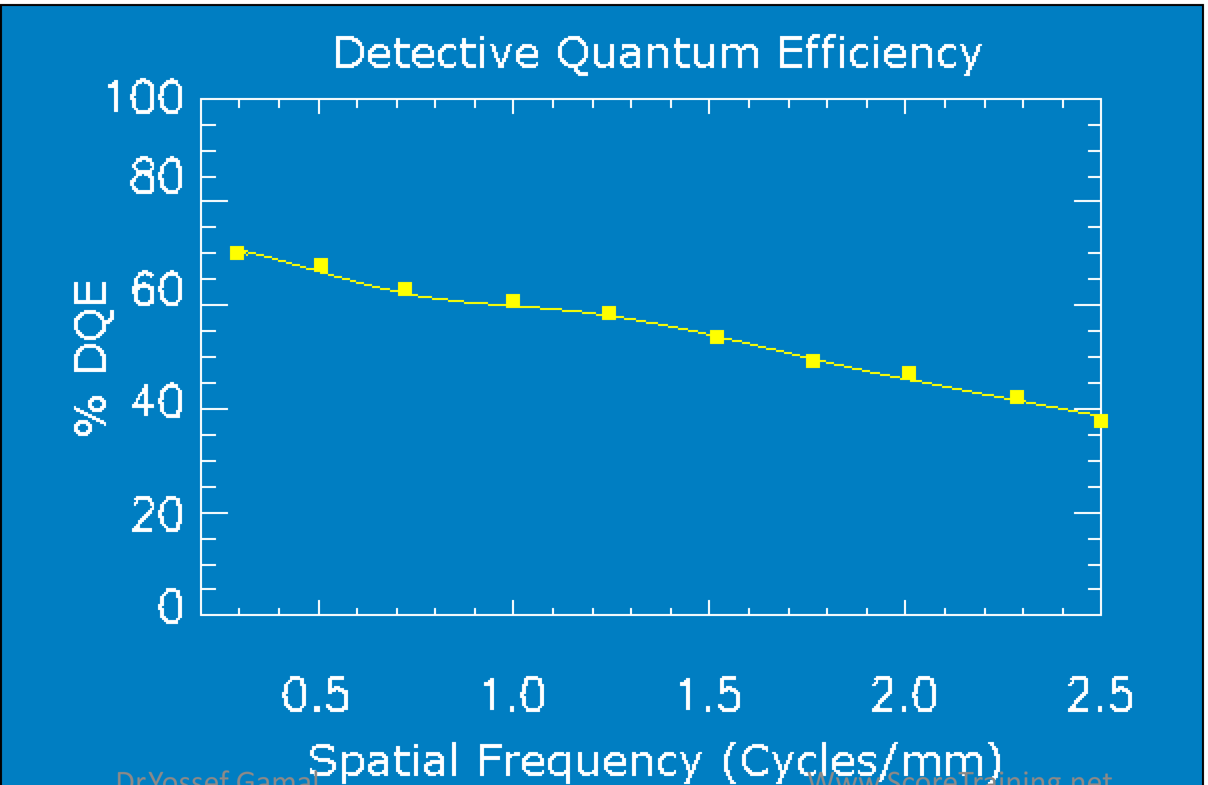
- If all X-ray photons are detected and no noise is added → DQE = 100%

- Factors affecting DQE:

1- Spatial frequency

↑spatial frequency → ↓DQE





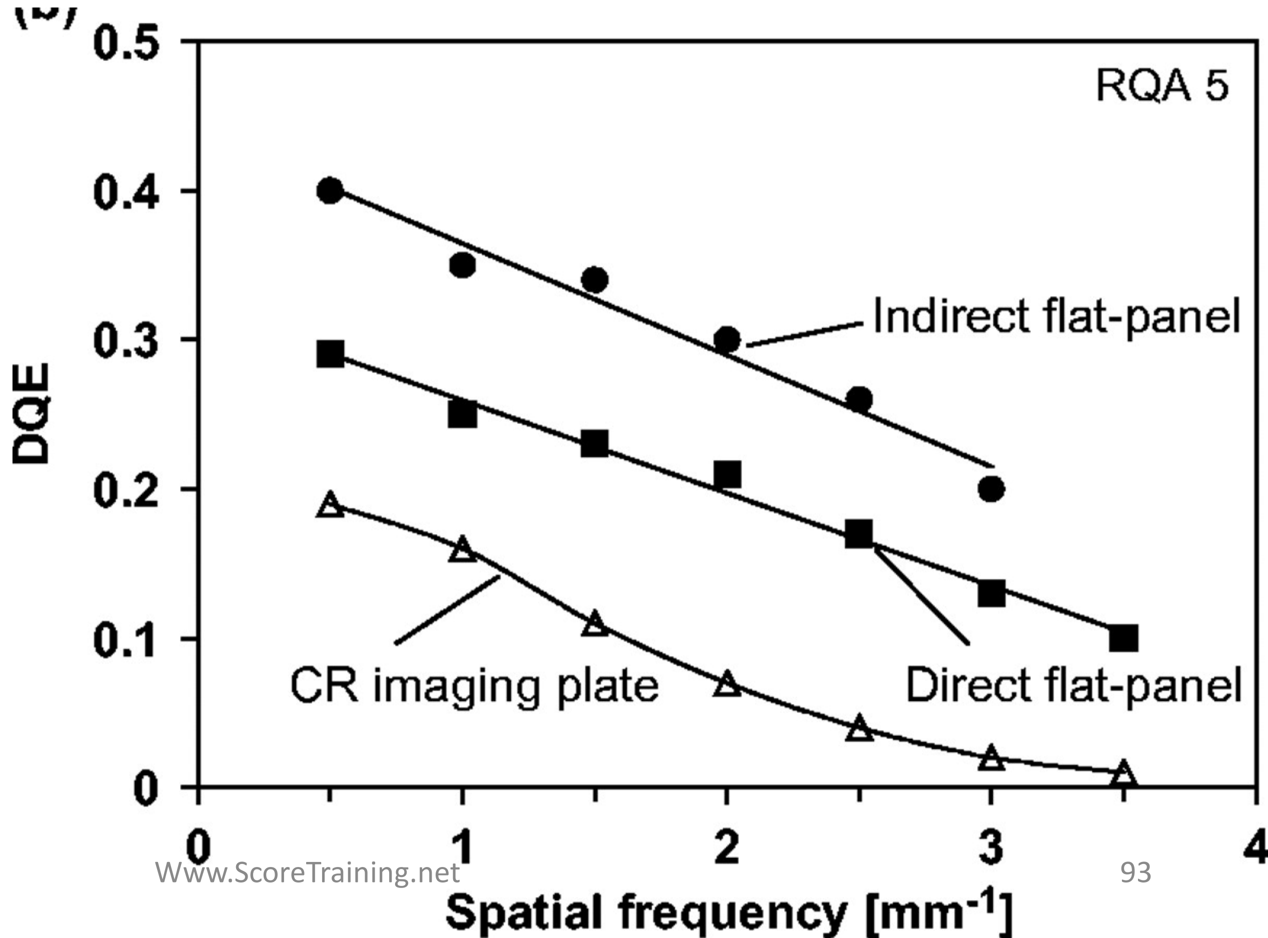
2) detector type:

- Depending on the absorption efficiency , structural noise and MTF of the detector
- Indirect DR › direct DR › CR
- CR and film screen has comparable DQE

N.B. Low selenium Z are the cause of moderate X-ray absorption and relatively low DQE (despite very high MTF)

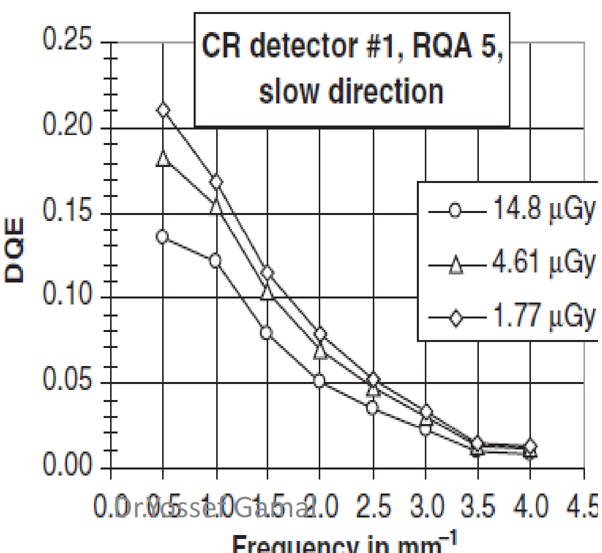
3) Fill factor:

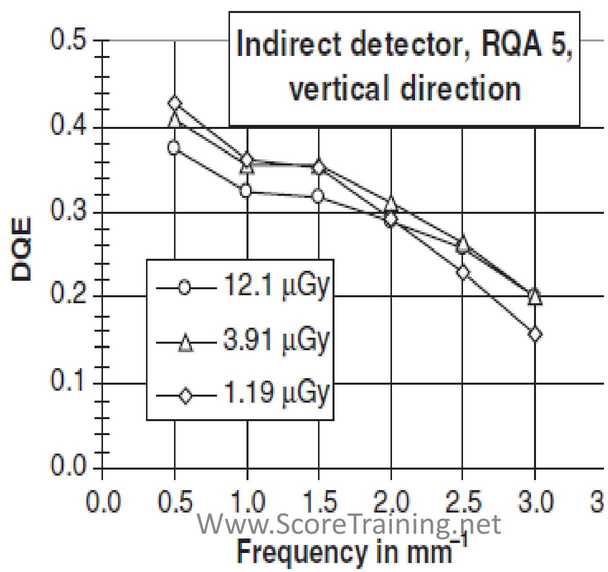
↑fill factor →↑fraction of absorbed photons →↑DQE

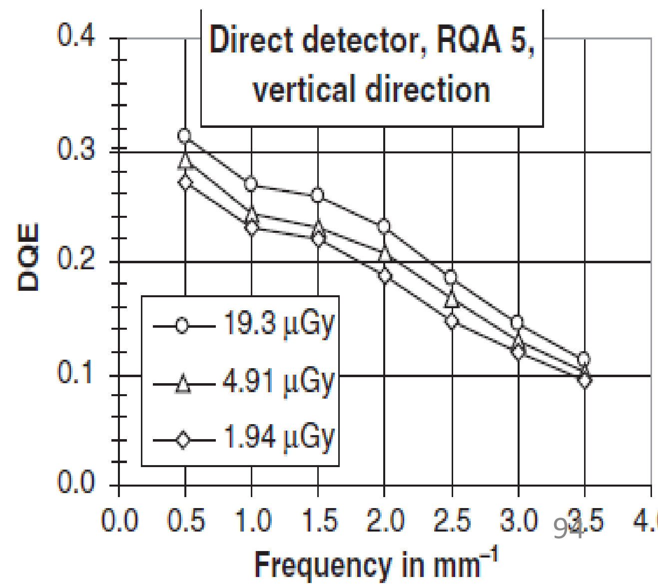


4) Exposure levels:

- The DQE of the CR detectors and is best for the low-exposure level.
- DQE of the indirect detector has no dependency on exposure level.
- The best DQEs of the direct detector is obtained at the highest exposure level





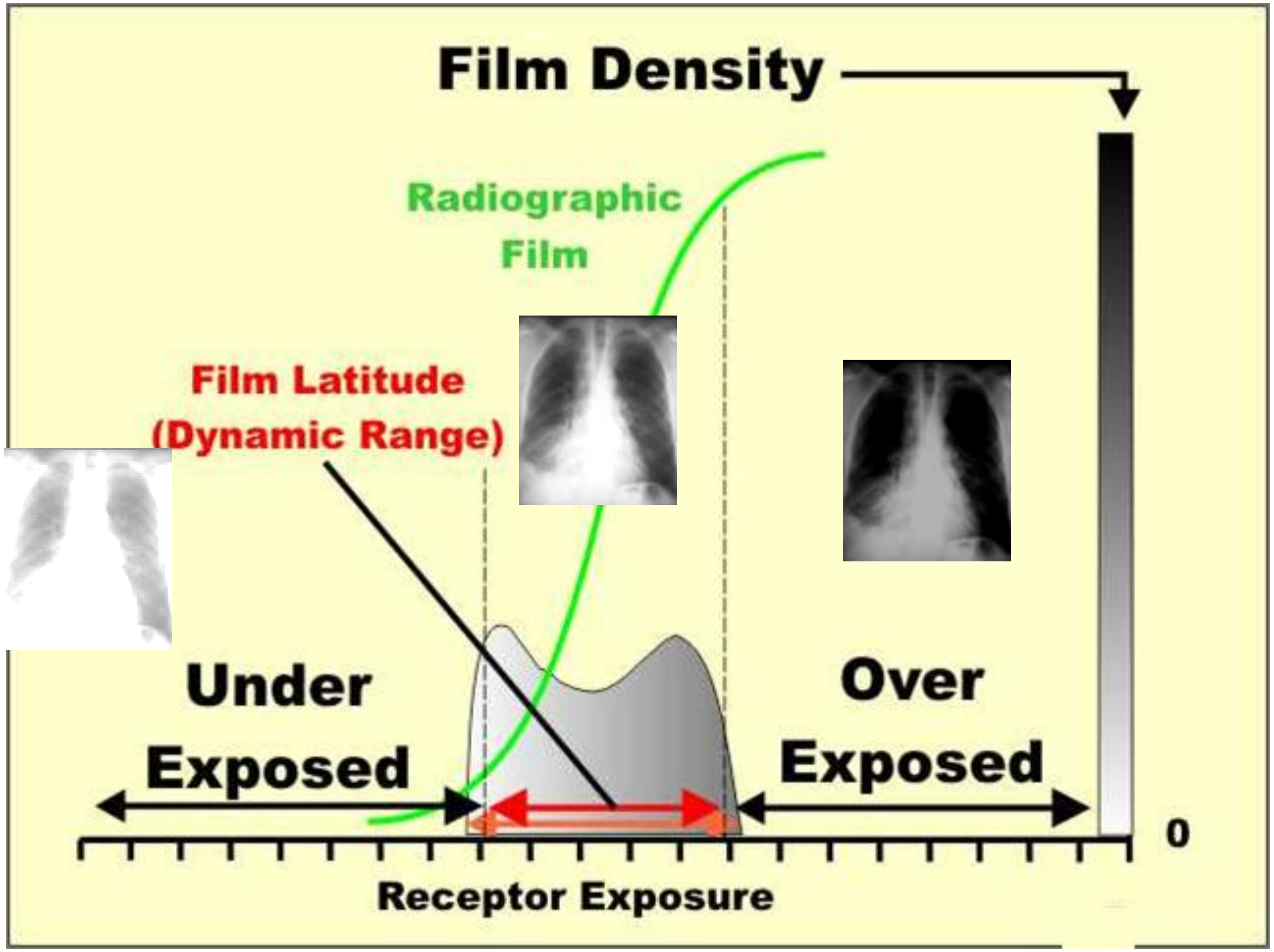


Radiation exposure with digital systems

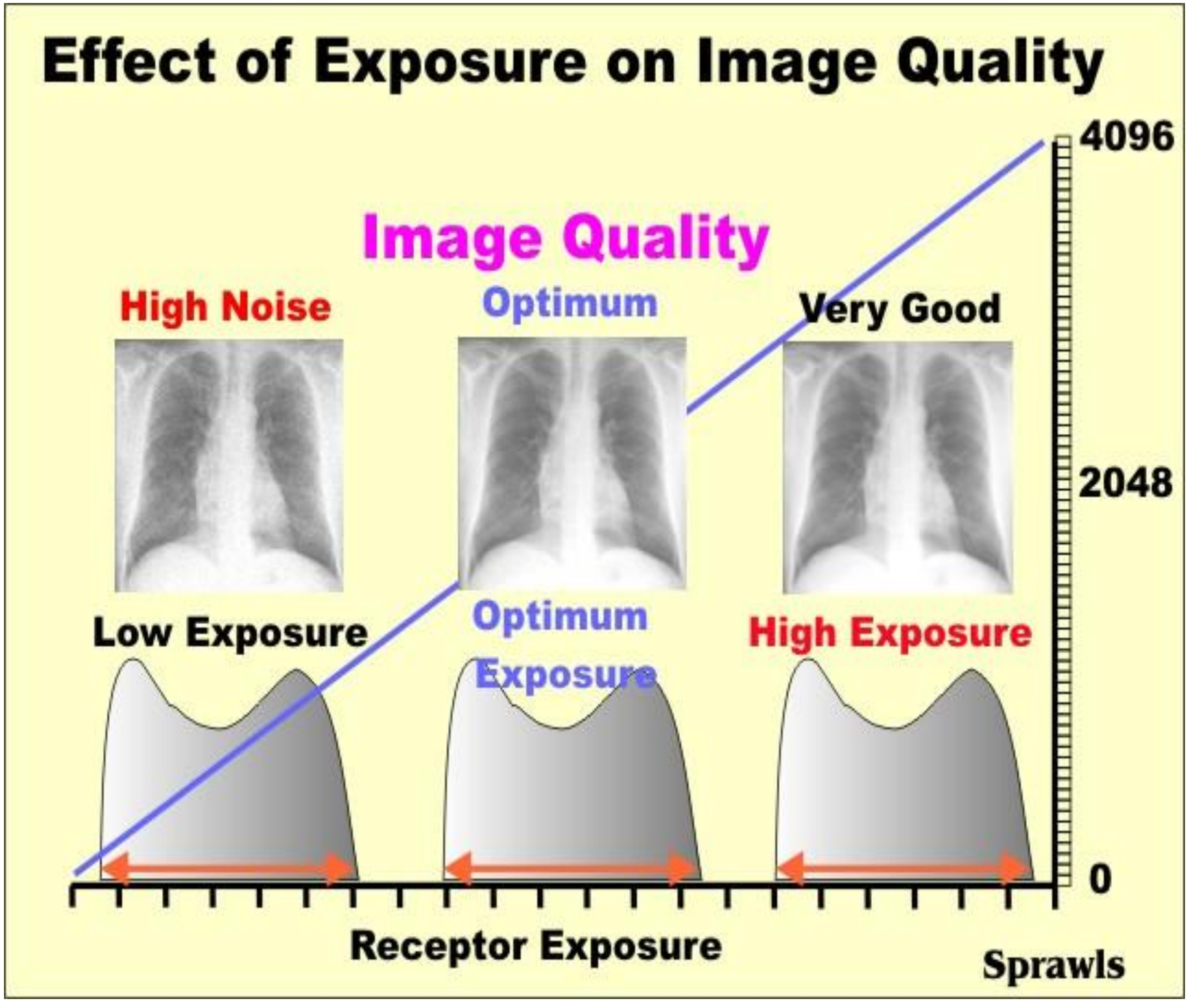
- Most of digital detectors decrease the exposure in comparison to film screen because:

1) ↑DQE → lower exposure for the same image quality (indirect DR has highest potential to decrease the exposure
2) Wide dynamic range → decreased number of failed exposures

Detector dose indicator



- Effect of wrong exposure in film screen system



- Effect of wrong exposure in digital system
- Almost impossible to under or overexpose CR / DR

Detector dose indicator (DDI) = exposure index

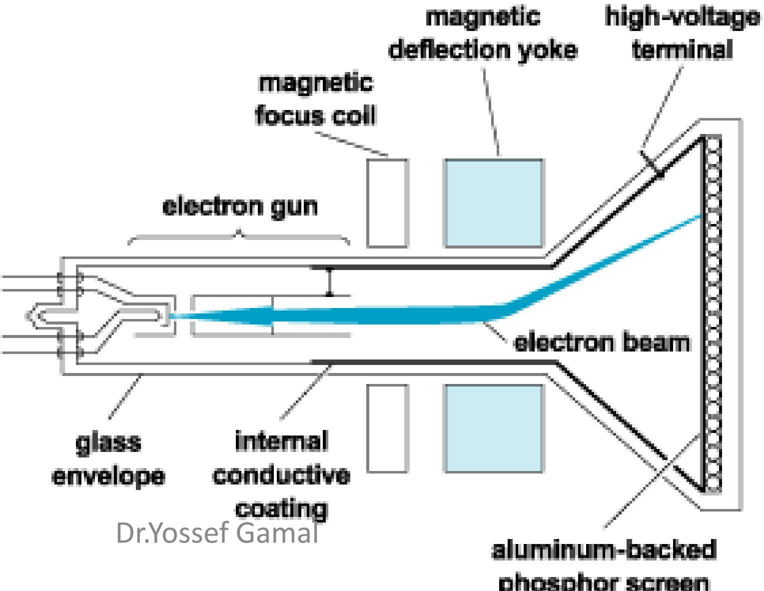
- Each manufacturer provides exposure feedback to technologist
- Displayed on CR reader monitor
- Displayed on DR workstations
- If DDI is higher than recommended range, patient is overexposed
- Manufacturers provide normal range of DDI which is exam dependant
- Some manufacturers give DDI inversely proportional to dose (↑ DDI = underexposure)
- Others give DDI = log exposure , i.e. doubling the dose = increase of DDI 15%

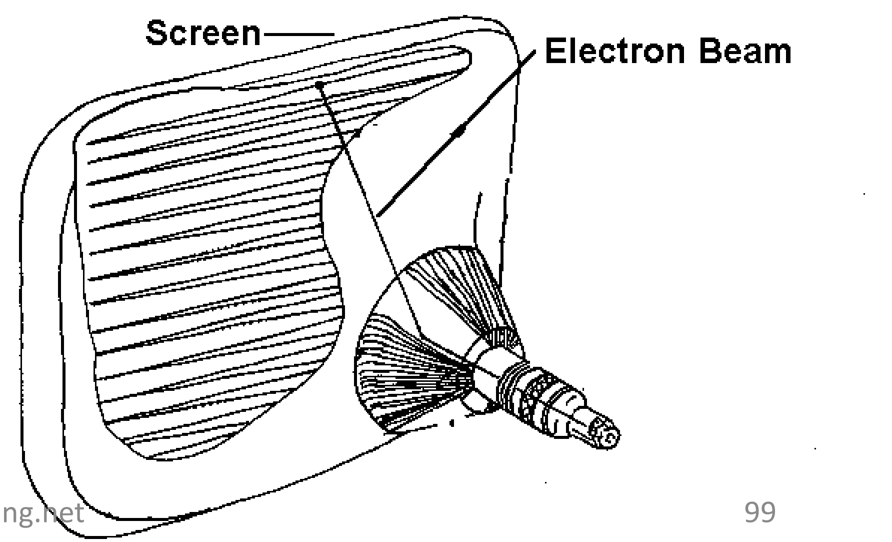
Receptor Exposure	Kodak EI	Fuji S Number
0.5	1700	400
1	2000	200
2	2300	100
4	2600	50

Soft copy viewing

A) Cathode ray tube (CRT)

- Scanning electron beam (its intensity is modulated according to pixel value)
- 1250 scan lines
- Resolution in the perpendicular plane of lines depends on the number of lines
- Resolution parallel to the plane of lines is limited to modulated signal frequency
- They have a big back and take up space on desk





B) Flat panel monitor (Liquid crystal display = LCD)

- Screen is formed of pixels (2500 x 2000)

i.e. double the lines of CRT

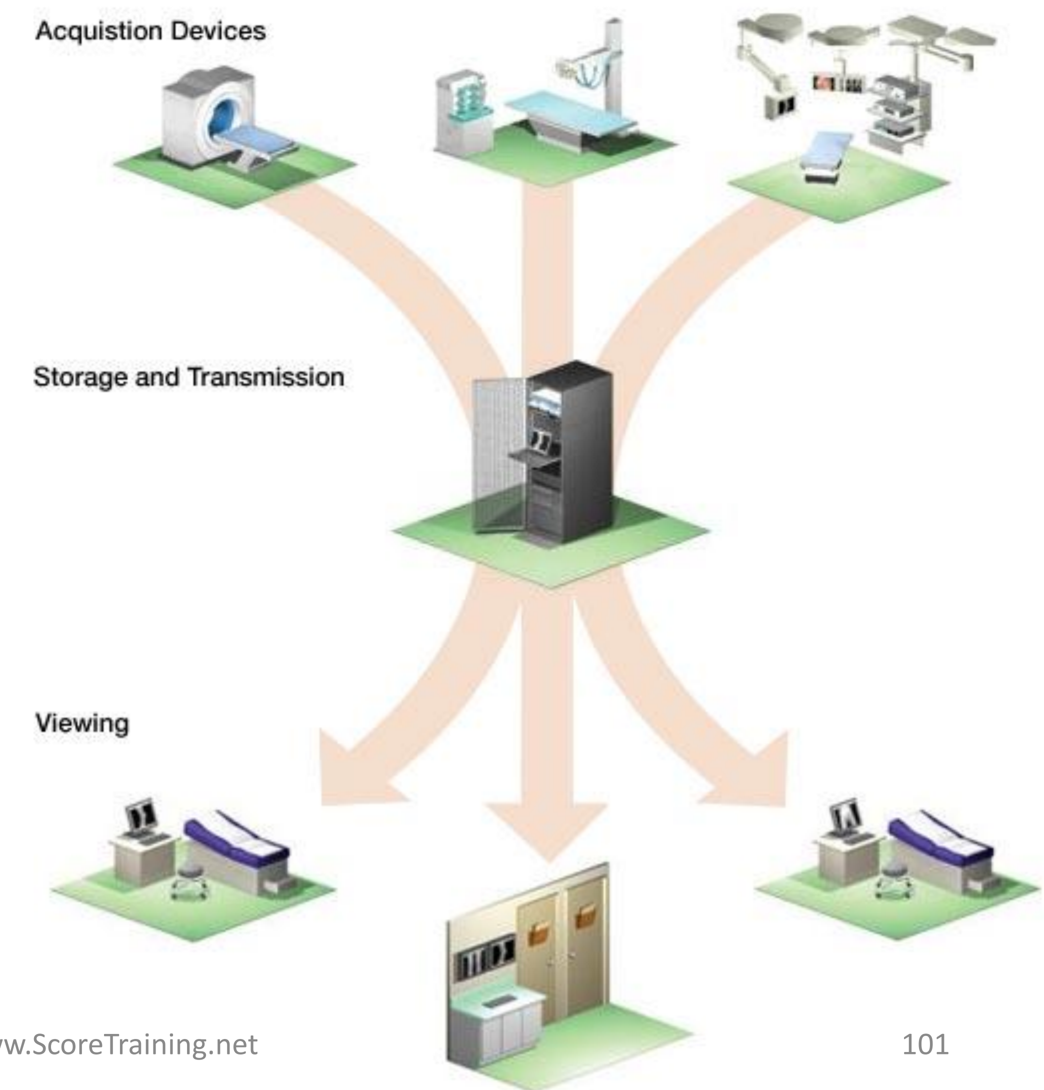
- Zero geometric distortion

PACS (Picture archiving and communication system)

- Networked group of computers, servers, and archives to store digital images
- Uses of PACS:

1) Provides image access to multiple users in multiple locations at the same time (not only in the hospital)
2) Secure storage of the images (with elimination of film stores)
3) Integration of the images with other electronic patient's records

- Disadvantage: Cost



- PACS workflow manager:

Computer in the heart of PACSs that controls flow of images and information (send images from and to the archive)

- Types of the archives:

1- short term: for current cases (e.g. inpatients) → rapid access on demand

2- long term: images that unlikely to be of immediate interest → access time is slower

Images are transferred between long term to short term automatically

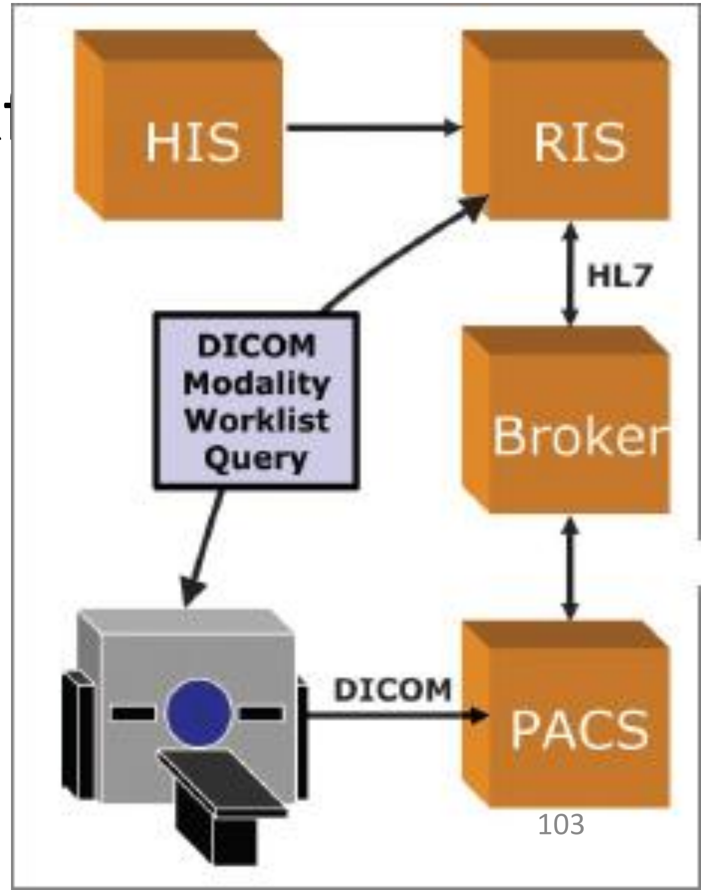
3- backup archive : for security

PACS is connected to other information systems:

1- hospital information system: patient demographics…etc.

2- radiology information system: appointments… etc.

PACS broker: interface between workt these systems



workstations that access the image:

1- Modality workstation: supplied with the image equipment , may be able to display images from archives

2- Reporting workstation: used by radiologist , must have high quality display monitors and software tools for image manipulation

-Display environment must be suitable: background lightening is kept low (extraneous light reduce film perceived contrast)

3- Review workstations: with lower quality monitors and limited software(reporting from these workstations must be done with caution)

N.B: Access to archives is available through web browsers (with passwords) , should not be used for diagnosis

Image file storage:

- Stored in the form of **DICOM** format
- Film file contains:

1- Basic digital data that allow display of the image

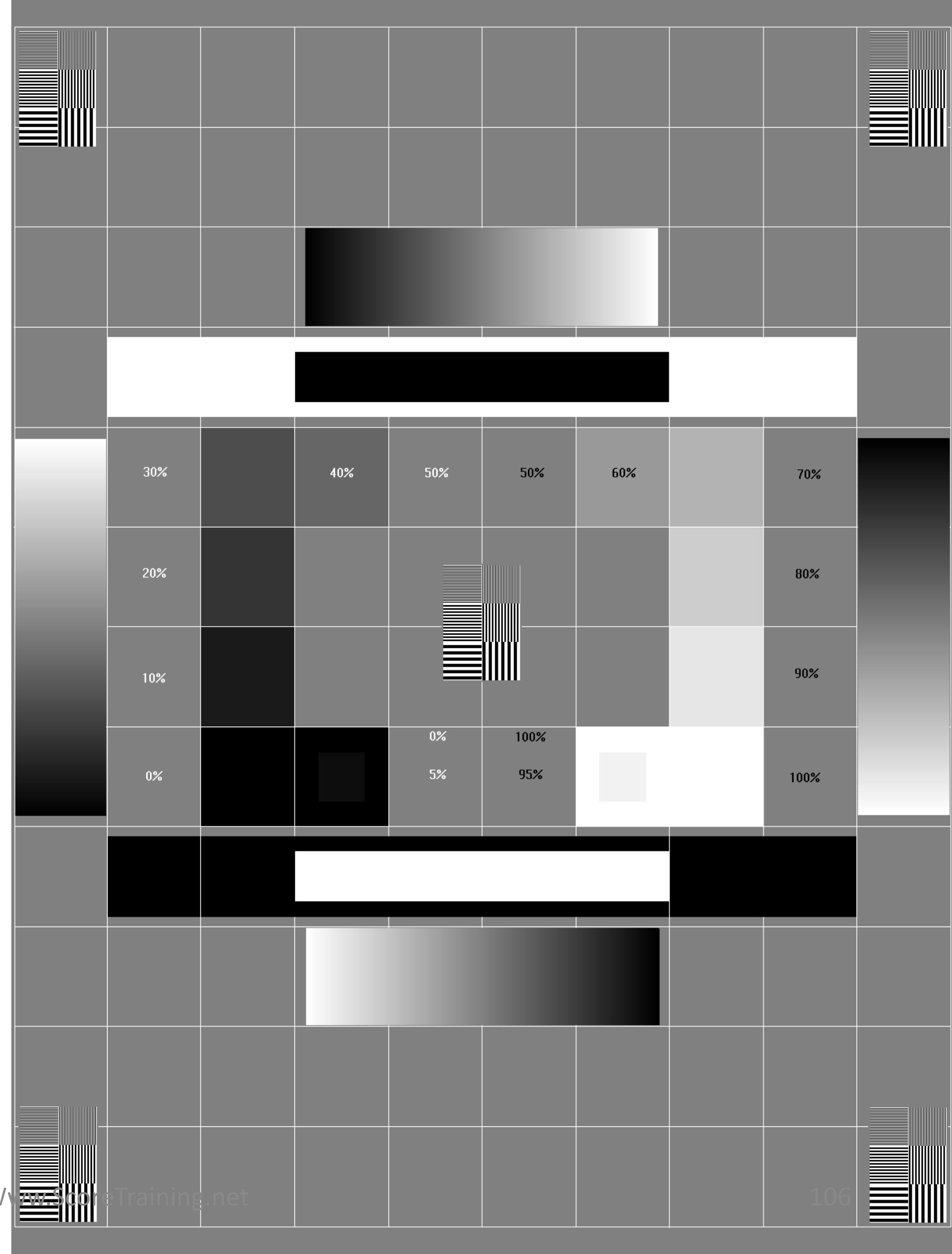
2- Annotations

3- Display preferences ….etc.

Quality assurance for display monitors

SMPTE test pattern:

- Areas of varying brightness from 0% to 100% in 10% steps
- Smaller (5%) contrast squares within zero and 100% squares (if not visible →monitor is out of calibration)
- Do not adjust the image using contrast and brightness controls




Thank You!